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DEVELOPMENT OF A WHOLE BODY FLYING-SPOT X-RAY MEDICAL UNIT.(U)

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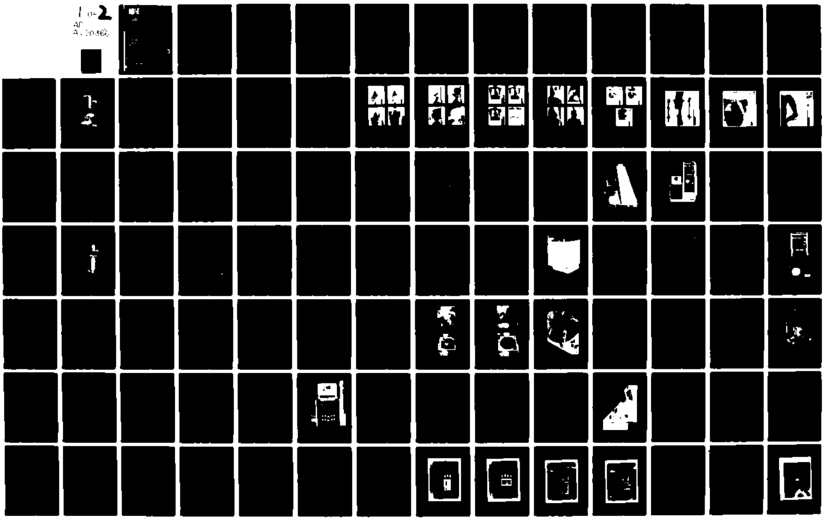

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FINAL SCIENTIFIC REPORT

DEVELOPMENT OF A WHOLE BODY FLYING-SPOT X-RAY MEDICAL UNIT

BY:

JAY A. STEIN

CONTRACT NO.
DAMD 17-74-C-4071

FOR THE PERIOD:
1 MARCH 1974 -
31 OCTOBER 1979

SUPPORTED BY:

U.S. ARMY MEDICAL RESEARCH
AND DEVELOPMENT COMMAND
FORT DETRICK, FREDERICK, MD. 21701

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20. ABSTRACT (Continue on reverse side if necessary and identify by block number)

The development of a flying spot X-ray scanner is reported. It is intended for field triage usage. Clinical trials were conducted at two sites, and the results are reported. This final report includes as an appendix the preliminary report of April 1975, which encompasses specifications for the scanning X-ray system.

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ILLUSTRATIONS

ILLUSTRATIONS

1.0 INTRODUCTION

This report reviews and summarizes a program that was initiated in 1974 to produce and evaluate a scanning X-ray system to screen combat casualties rapidly, conveniently, and with a low dose.

It has gone from a breadboard analogue system to a prototype digital system placed in the field for clinical evaluation.

A historical perspective is given in Section 2.0 which traces the development in detail. Section 3.0 gives a system description of the final system evolved under this contract. Section 4.0 discusses the clinical trials undertaken and their results.

2.0 HISTORICAL PERSPECTIVE

The subject contract (DAMD 17-74-C-4071) was let by the U.S. Army Medical Research and Development Command to American Science and Engineering (AS&E) in March 1974.

The goal of this program was to develop a laboratory breadboard whole body flying-spot X-ray imaging system suitable for the medical evaluation of diagnostic image quality for applications in military medicine. The system utilized the flying-spot technology developed by AS&E for parcel inspection.

Figures A1-2 and A2-2 show an X-ray tube which serves as a source of X-rays. A simple slit collimator is used to create a narrow fan beam of X-rays. The fan beam of X-rays is, in turn, modulated by a rotating disc into which a series of slits have been cut. As the collimating disc rotates, a small pencil beam of X-rays moves across the object under examination. The cross-sectional area of the pencil beam forms a flying X-ray spot. This pencil beam of X-rays is partially attenuated by the object being examined with subsequent detection of the transmitted portion of the beam taking place in a rod-shaped scintillation detector. This detector is thallium-activated sodium iodide (NaI(Tl)) which detects virtually 100 percent of the transmitted portion of the pencil X-ray beam.

The NaI(Tl) crystal is viewed by a photomultiplier tube (PMT) which detects the light output of the NaI(Tl) and converts it to an electrical signal. This signal is analogous to the video signal in a television system. During one scan of the pencil beam from one end of the detector to the other a line image is produced.

The second dimension of the video image is generated by moving the object to be imaged with respect to the source-collimator-detector

system so that the motion causes each subsequent scan of the pencil beam to be slightly displaced from the previous one. If the line images of each of these successive scans are displayed one after another, the result is a complete two-dimensional image of the region under examination. This display is again analogous to the pattern on a video monitor which displays pictorial information as a series of many scan lines. The use of television display techniques is, in fact, well suited to the system since the output of the detector is completely analogous to an ordinary video signal. The same techniques which are used for storing and displaying video signals are used to store and display the radiological images which are generated with the system.

Figures A1-3 and A1-4 are photographs of the entire system with its longitudinal patient transport. The lateral scanning mechanism is housed beneath the table. The unique features of this type of scanning X-ray imaging system are:

a. The high detection efficiency of the X-ray detector results in an exceedingly low radiation dose to the subject examined.

b. The effects of scattered radiation which degrade contrast in a conventional system are eliminated.

c. The image information is stored so that exposure time is limited to that period required to scan the area of interest once (about 10-15 seconds).

d. The image is available for examination instantaneously after completion of the scan.

e. The image signal can be processed in several ways, including a high resolution television display, a hard-copy photographic print, or with remote transmission equipment.

f. There is no radiation hazard to the operator from stray radiation, which allows an operator to remain with the patient during the examination.

g. Specially shielded rooms or other facilities are not required.

h. The entire body, or any part of it, may be examined readily.

The system proved to be very successful in meeting its goals and produced pictures of moderate quality as shown in figures A5-9 through A5-14.

A complete report of this phase (Phase I) of the program is included in Appendix A.

In the next phase (Phase II) AS&E developed a prototype version of the system featuring the same intrinsic technology, improved circuitry and a moving source and detection system. Very early in its development, however, it was realized that the basic limitation on the picture was in the analog storage of the image. Noise in the image contributed by the storage medium exceeded the basic noise in the X-ray photon statistics. General principles suggest that the X-ray photon noise must be the limit to the picture noise for maximum information. To improve this situation, AS&E decided to develop and install a digital image storage and display system. This development and equipment was paid for by AS&E.

This made a dramatic change in the picture quality and allowed the utilization of the available dynamic range.

This system was then placed in clinical trials from September 1 to December 31, 1977 at Children's Hospital Medical Center in Boston. It was then placed at the Maryland Institute for Emergency Medicine (MIEM) for clinical work from February 1978 through October 1978. The unit was moved from MIEM to Fort Detrick, Frederick, Maryland.

3.0 SYSTEM DESCRIPTION

3.1 Scanner

Referring to Figure 3-1 the output of a conventional rotating anode X-ray tube, (a) is collimated by a tungsten slit, (b) into a narrow fan beam. The fan beam is further collimated by a series of tungsten slits, (c) arranged radially on a lead-filled aluminum chopper wheel. The lead and tungsten fully attenuate X-rays except in the overlap of the slits. The motion of the wheel causes the slits to traverse the fan beam repeatedly, generating a scanning pencil beam of X-rays, (d).

The pencil X-ray beam is partially attenuated by the subject and the unattenuated X-rays are absorbed by the photon detector, (e). The pencil beam of X-rays moves along the length of the patient due to a transverse motion of the entire assembly including the X-ray source, the chopper wheel, and the detector. The patient remains stationary.

The detector is a 28-inch long NaI(Tl) scintillation crystal coupled to a photomultiplier, and nearly 100% of the X-rays which are not attenuated by the patient are detected. The electrical signals obtained at the output of the photomultiplier are pulses, the amplitude of each pulse being proportional to the energy of a single detected X-ray photon. Since the rate of X-ray photons incident on the detector is large, these pulses add together giving a net signal which at any instant of time is proportional to the incident X-ray flux in the attenuated X-ray pencil beam.

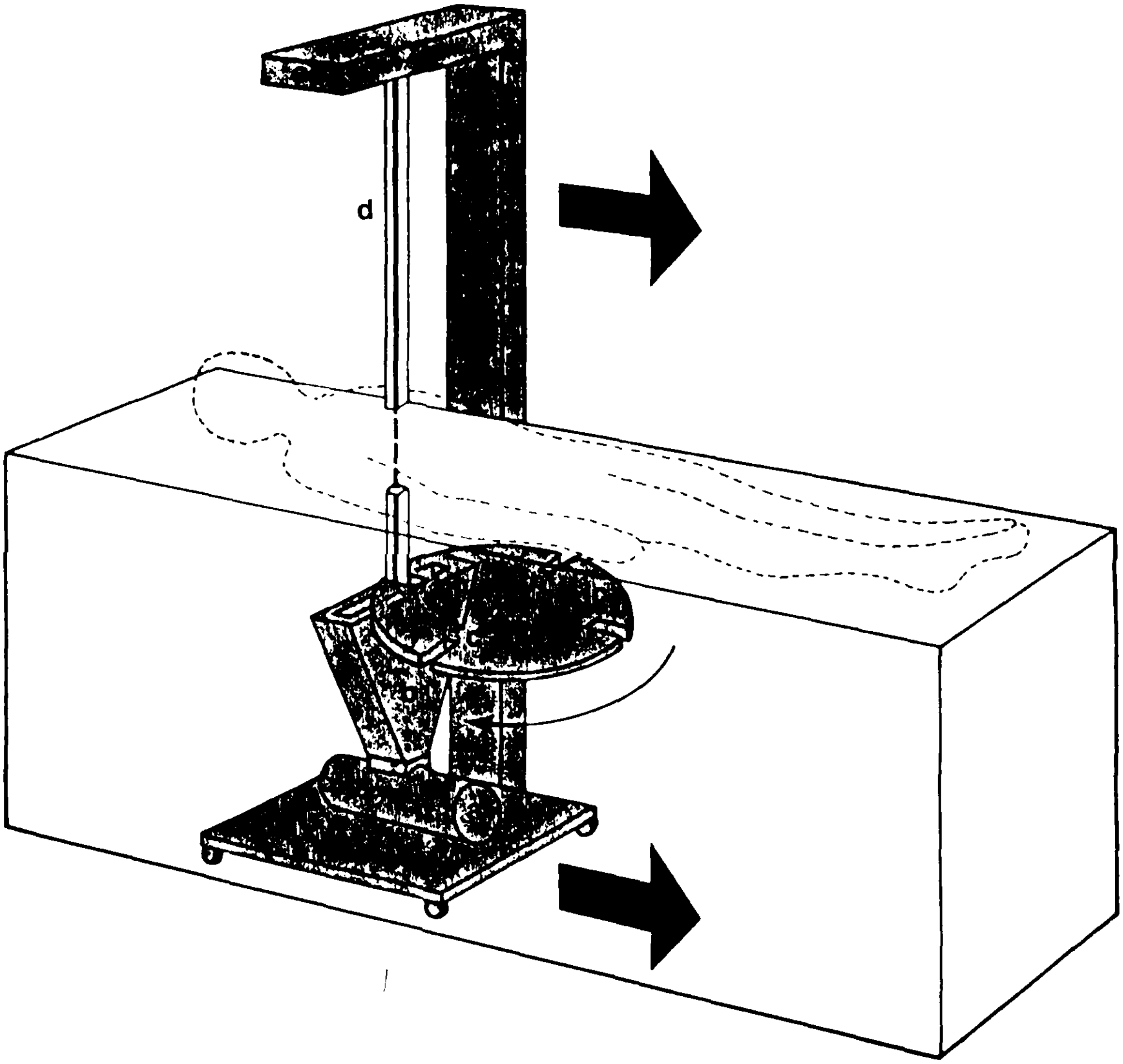

Figure 3-1. Principle of Operation

The electrical signal from the detector, during one scan of the pencil beam from one end of the detector to the other, corresponds to a one-dimensional radiographic "line image" of the object. This line image is analogous to one scan line on an ordinary television monitor. The second dimension of the image is generated by virtue of the motion of the source-collimator-detector plane with respect to the patient. The series of line images is sequentially stored in digital form.

After the 15-second X-ray exposure is complete, the radiographic data are read out line-by-line onto a television monitor. The readout is sequentially ordered in the same manner in which the data are read into storage so that the image on the monitor screen is the X-ray shadowgraph of the subject.

The scanner's chopper wheel shown in Figure 3-1 has three sets of radially mounted slits, each of which is associated with a particular examination field size. The large field is approximately 15 inches wide by 20 inches long. The medium field is approximately 6 inches by 8 inches and the small field is 1.5 inches by 2 inches. These modes are selected by three front panel buttons which control automatic circuits which position the scanner and slot, and electrically select proper synchronizing signals. The result is that the operator may "zoom" in on an area of particular interest and examine it in greater detail, with much better definition.

3.2 Display System

The detector information is processed and stored by means of a mini-computer. By this method, the video signal is accurately digitized and stored; stored data in digital form allows a radiograph to be processed and displayed an unlimited number of times without any deterioration in the original image. The physician then has the ability to select a particular change in density which then can be amplified in contrast at his convenience.

The basic computer processing of data is the accurate specification of the window width of contrast (gray scale expansion) and the window center level of contrast (median X-ray attenuation displayed). This ability to display selected contrast windows allows the presentation of any dynamic range in X-ray intensities using a series of necessarily limited gray scale displays. This overcomes the problem of taking a series of X-ray exposures as may be required for film radiographs.

3.3 Features

The type of scanning X-ray imaging system shown in Figure 3-1 overcomes many of the limitations of the conventional screen/film system. Much of the scattered radiation is automatically eliminated without the use of special equipment. This is illustrated in Figure 3-1, where it is clear that due to the very narrow X-ray detector aperture, only a small fraction of the total scattered radiation which would be incident upon a conventional screen/film system is detected, and only this much contributes to the background level.

As previously described, the use of NaI(Tl) crystal as a detector in a scanning X-ray imaging system results in nearly 100% X-ray detection efficiency and therefore minimizes the dose required for a given examination. More importantly, because the scintillation

detector is essentially noise-free (individual X-ray photons can be detected above the noise in the photomultiplier), there is no reduction in contrast due to a noisy detector such as film. Indeed, all of the noise is due only to photon statistics. Further, the dynamic range of the scintillation detector is at least 4 orders of magnitude so that the range of intensities that can be recorded is in excess of 100 times that possible with film.

4.0 CLINICAL TRIALS

4.1 Children's Hospital Medical Center (CHMC)

The unit was placed at CHMC from September 1 to December 31, 1977. This hospital was chosen for its proximity to AS&E. This allowed AS&E to become familiar with logistic and maintenance problems in a location near their main office. CHMC was interested in these trials because of the new technology involved and the low dose supplied to the patients.

The protocol adopted was to perform parallel studies utilizing both normal film screen techniques and Micro-Dose techniques. A few of the many images are shown in Figure 4-1 to 4-5. These were chosen to demonstrate the machine's flexibility, scatter rejection, and capability for post data processing. While CHMC was able to demonstrate the normal range of radiological studies, they lacked access to the serious trauma cases of interest for the Army's application.

4.2 Maryland Institute for Emergency Medicine

For this reason the unit was moved to the Maryland Institute for Emergency Medicine (MIEM). This unit deals solely with the seriously traumatized patient. MIEM operates a helicopter service to pick up patient at the scene of an accident. The helicopters land on the roof of the unit and the patients are brought to the operating floor level via elevator. The Micro-Dose unit was located between the elevator and the operating suite with a remote display in the suite. The patients were put on the machine while on their stretchers. Therefore the delay was minimized. In fact, only about an additional 30 seconds were required to obtain the images. However, because of the remote terminal, the surgeons could review the radiographic images before the patient arrived at the suite. This was found to be a very useful feature. Figures 4-6 through 4-8 show a few of the many images obtained there.

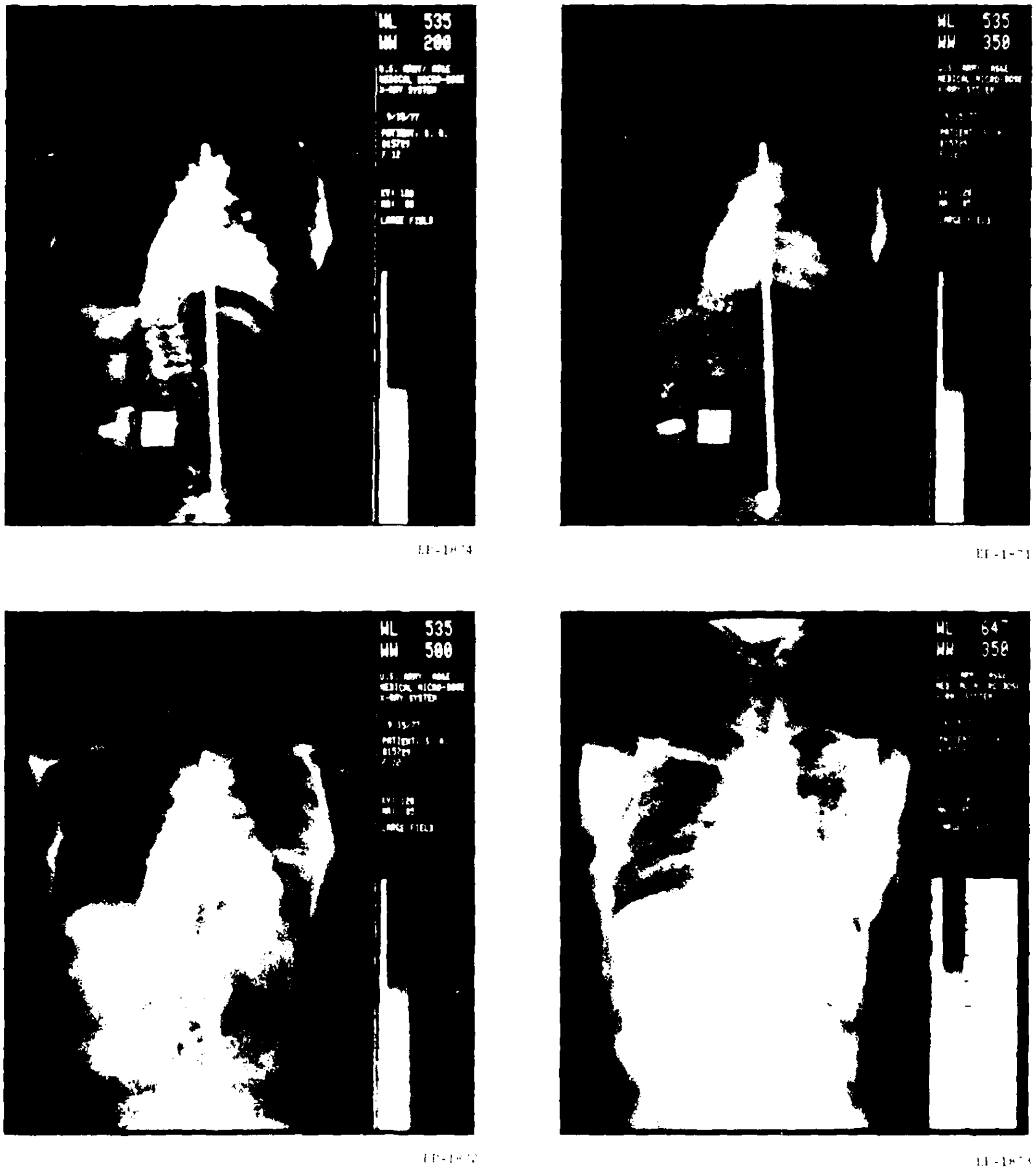

Figure 4-1. Four radiographic images from a single PA exposure (0.4 mR) of a 12 year old female in a body cast. Three of the displays were set to a fixed window level (WL), but a variable window width (WW). The right hand images are set to a fixed window width, but at different levels.

Figure 4-2. Four nasopharyngographic displays of a 13 year old male, due to a single 0.6 mR exposure. The lower two images are identical except for a gray scale reversal, instantly available by the push of a control button.

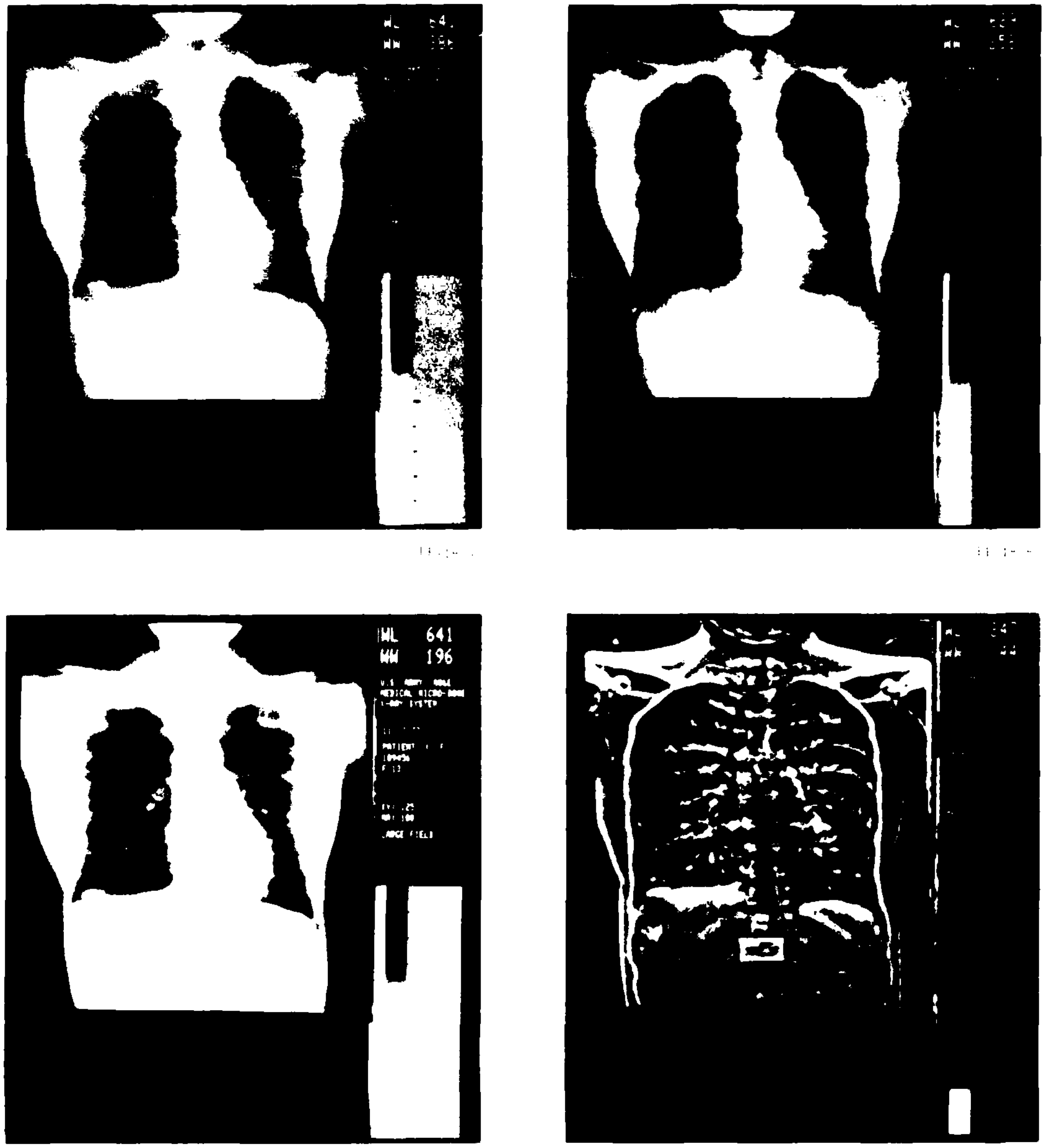

Figure 4-3. Four images of a 13 year old female, due to a single PA projection taken at 0.4 mR. One image has been processed by removing the low frequency components of the stored image data. This particular processed image was useful to this patient's surgeon because of the convenient localization of the two cancerous regions with respect to the ribs, which are all visible due to the processing.

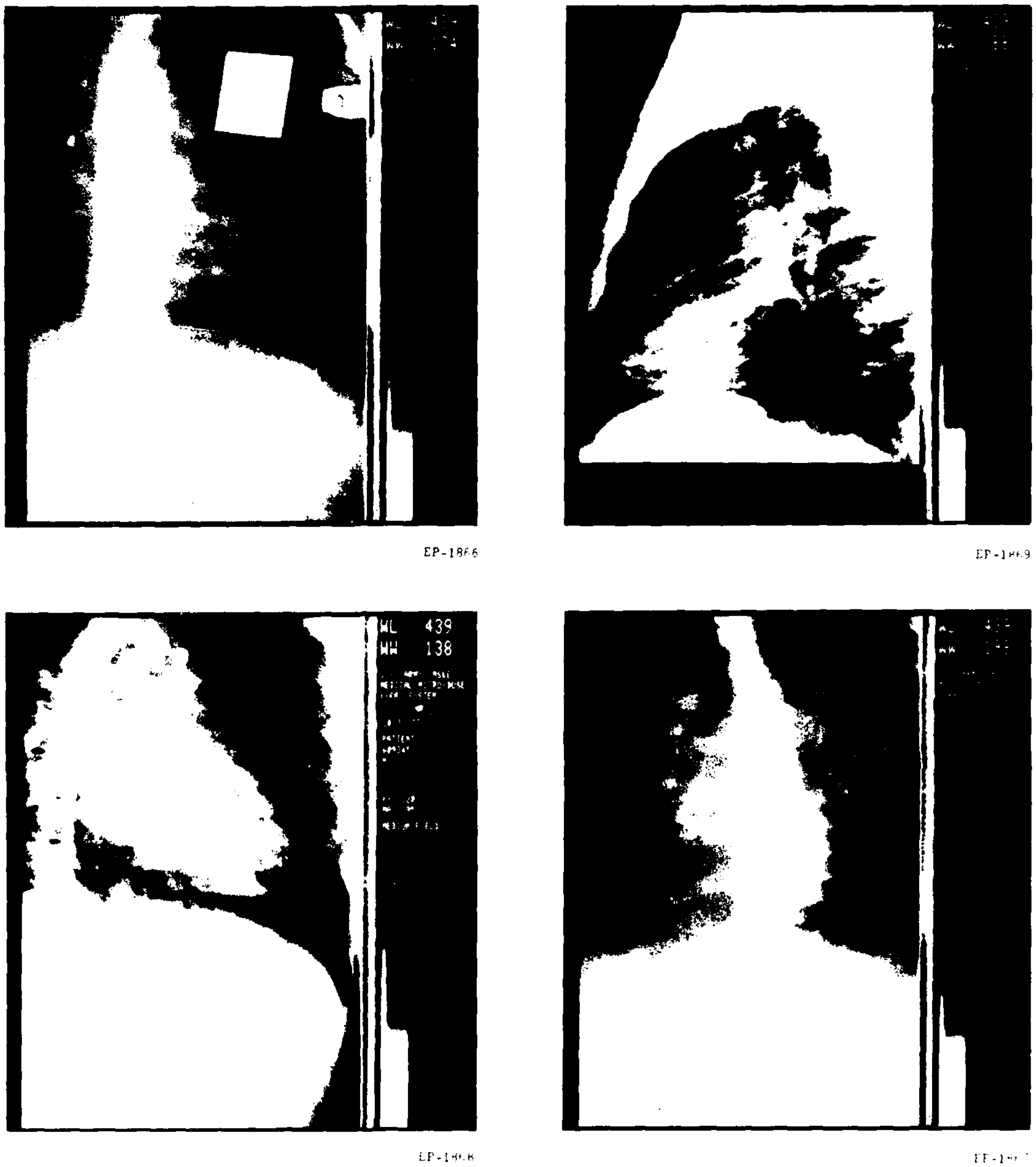

Figure 4-4. A dynamic cardiac study of an 11 year old male using a PA, lateral, RPO and LPO projection. Each of the four exposures were taken at 0.4 mR. Because each radiographic image is created by high speed transverse sweeps of an X-ray pencil beam (0.03 sec/sweep), conveniently obtained kymographs are generated, a feature which is unique to this system.

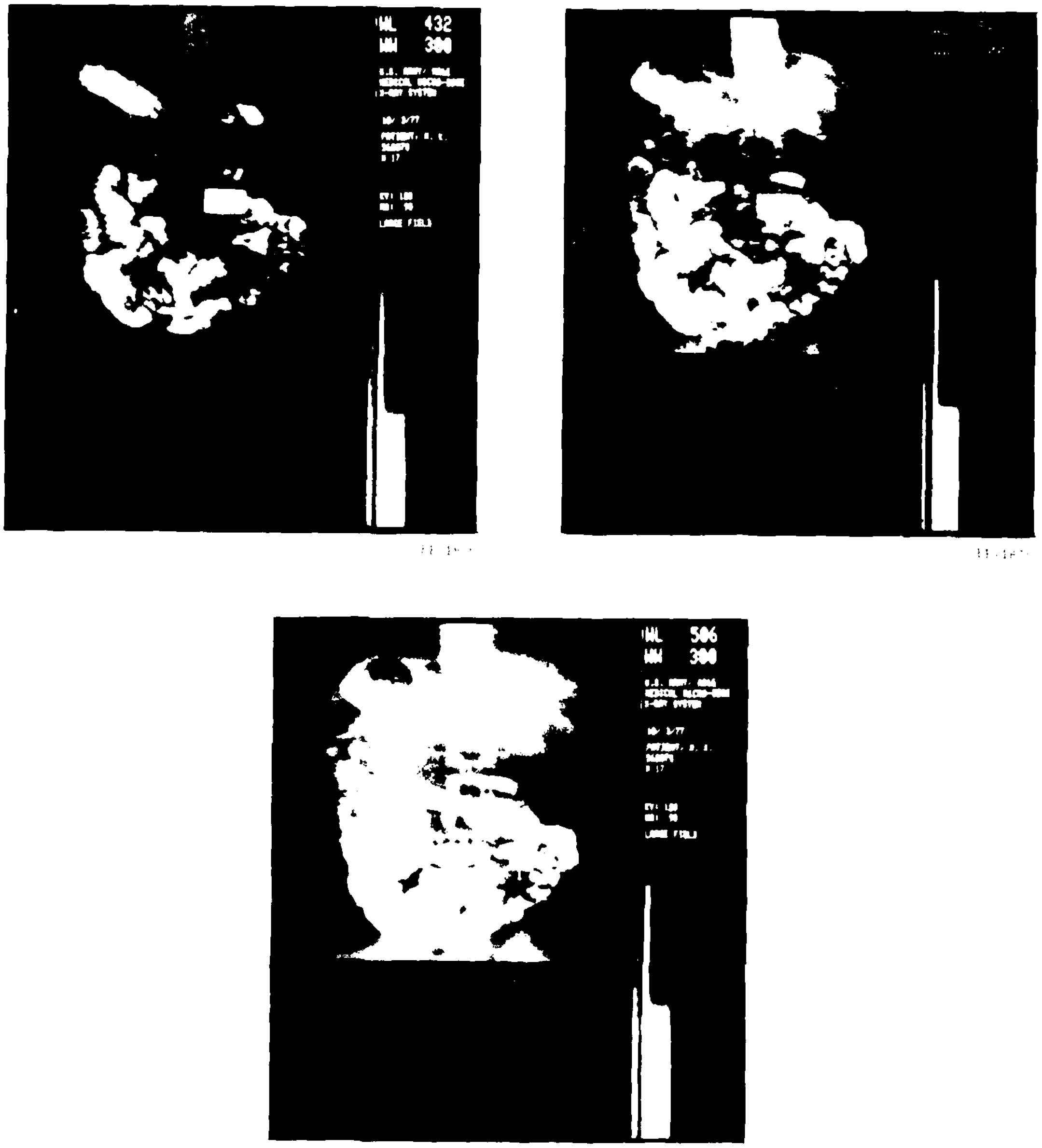

Figure 4-5. Three radiographic images, due to a single AP projection (prone position) of a 17 year old male taken at 0.4 mR. Barium sulfate was used in this routine gastrointestinal examination.

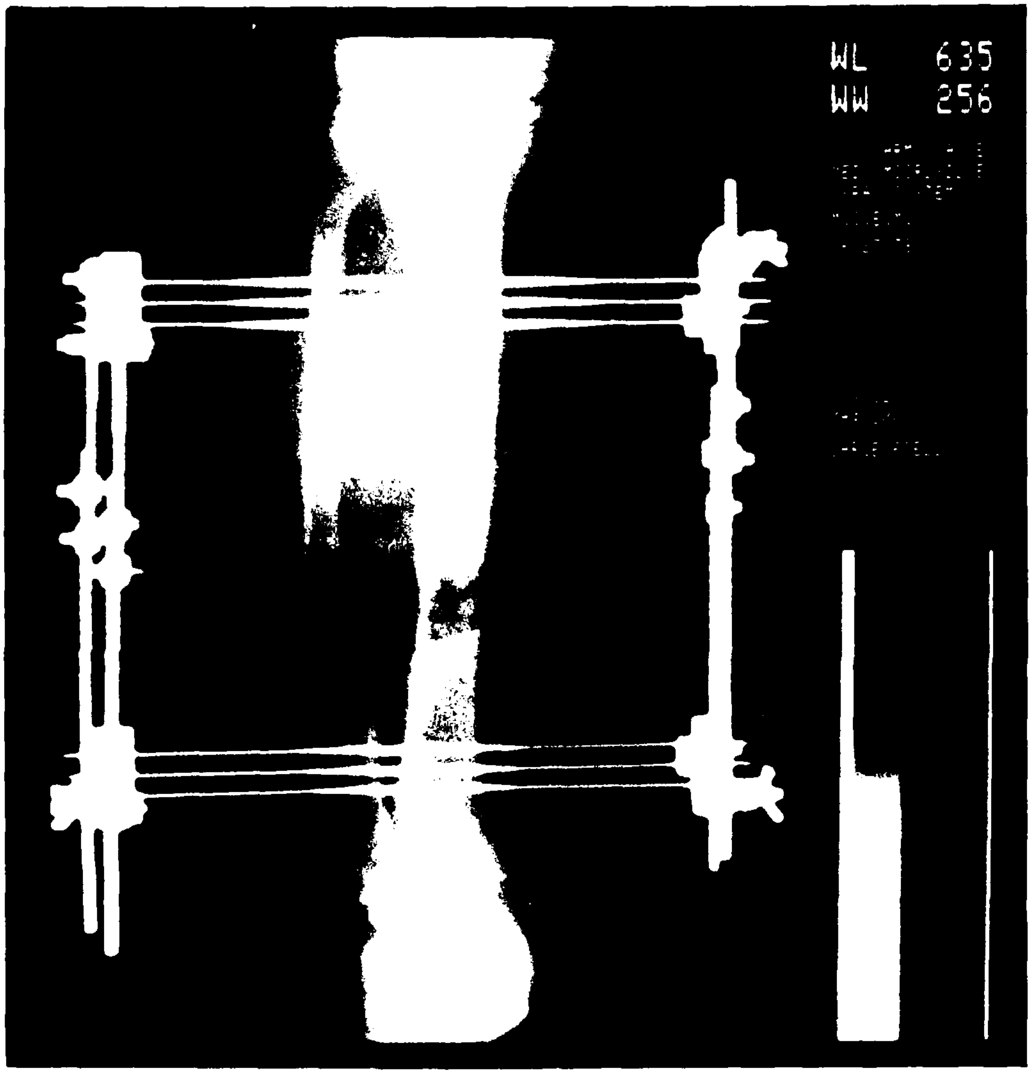

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Figure 4-6. A typical follow up exam from MIEM, showing broken bones stabilized with an external fixation device.

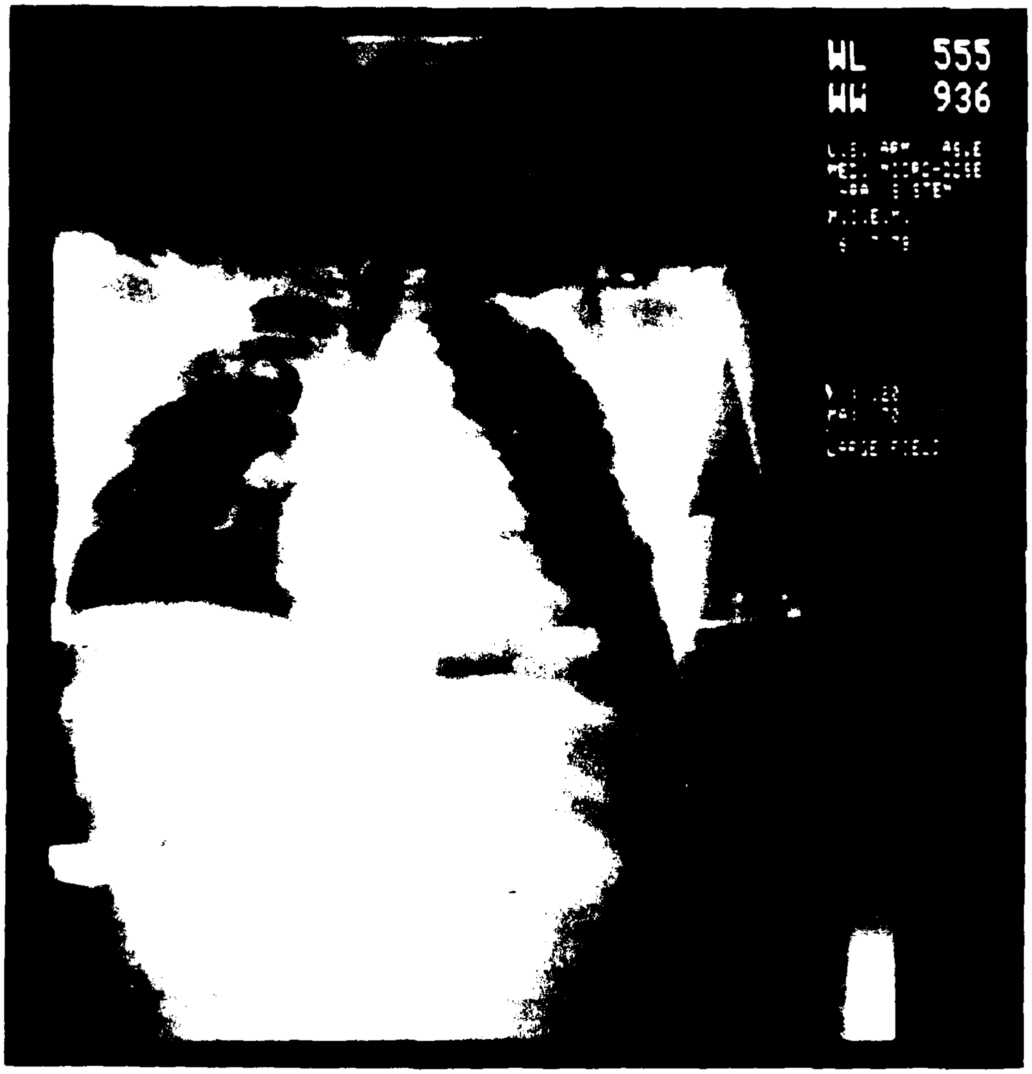

Figure 4-7. A typical admission exam from MIEM. The external structure shown overlain on the abdomen is a stretcher strut.

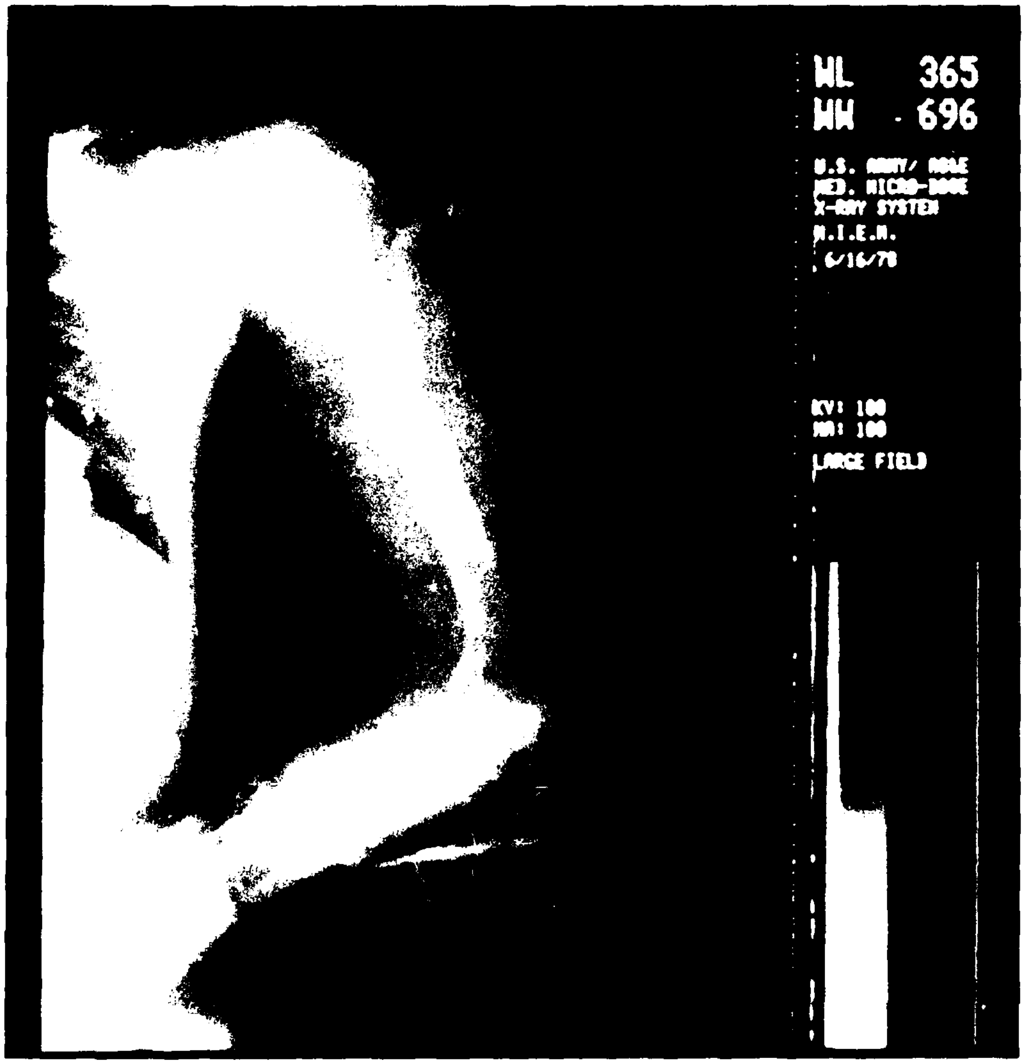

Figure 4-8. A typical follow up study from MIEM through plaster.

Some of the scientific output from these efforts have been:

1. Two "work in progress" papers presented at the 1977 meeting of the Radiological Society of North America. Presented by Dr. A. Rosenbaum and Dr. E. Frederick, respectively. These were based on the CHMC experience.

2. One paper presented at the 1978 meeting of the Radiological Society of North America by Dr. D. Curtis. This was based on the experience at MIEM.

3. Two scientific exhibits at the 1978 meeting of the Radiological Society of North America.

APPENDIX

DEVELOPMENT OF A WHOLE BODY FLYING-SPOT X-RAY MEDICAL UNIT

SCIENTIFIC REPORT

Jay A. Stein
and
Daniel J. Healey

April 1975

Supported by:

U.S. Army Medical Research and Development Command
Fort Detrick, Frederick, MD 21701

Contract No. DAMD 17-74-C-4071

American Science and Engineering, Inc.
955 Massachusetts Avenue
Cambridge, MA 02193

CONTENTS

ILLUSTRATIONS

ILLUSTRATIONS

SUMMARY

The objective of the program was to evaluate the image quality achievable with a low dose scanning X-ray imaging system designed specifically for the rapid and convenient whole body X-ray screening of combat casualties. A laboratory scanning X-ray imaging system was developed and X-ray images obtained under optimum operating conditions.

The X-ray images obtained with the system are of sufficiently high quality that they may easily be used to detect the presence of small bits of metal, such as shrapnel, throughout the entire body. Furthermore, there is good reason to believe that much more clinically useful information can be obtained. The high quality of the images and the verification that they can be displayed instantaneously in a medically meaningful manner indicates that a more extensive clinical evluation of this technique is justified.

1.0 INTRODUCTION

The purpose of this program was to design and construct a laboratory scanning X-ray imaging system and to evaluate the image quality obtainable with this low dosage technique. The system consists of a complete X-ray scanner and display unit as well as a transport apparatus for movement of the subject. The basic principle of operation is shown in Figure 1-1 and a block diagram of the system is shown in Figure 1-2.

An X-ray tube serves as a source of X-rays and a simple slit collimator is used to create a narrow fan beam of X-rays. The fan beam of X-rays is, in turn, modulated by a rotating disc into which a series of slits have been cut. As the collimating disc rotates, a small bencil beam of X-rays moves across the object under examination. The cross-sectional area of the pencil beam forms a flying X-ray spot. This pencil beam of X-rays is partially attenuated by the object being examined with subsequent detection of the transmitted portion of the beam taking place in a rod-shaped scintillation detector. This detector is thallium-activated sodium iodide (NaI(Tl)) which detects virtually 100 percent of the transmitted portion of the pencil X-ray beam.

The NaI (Tl) crystal is viewed by a photomultiplier tube (PMT) which detects the light output of the NaI(Tl) and converts it to an electrical signal. This signal is analogous to the video signal in a television system. During one scan of the pencil beam from one end of the detector to the other a line image is produced.

The second dimension of the video image is generated by moving the object to be imaged with respect to the source-collimator-detector system so that the motion causes each subsequent scan of the pencil beam to be slightly displaced from the previous one.

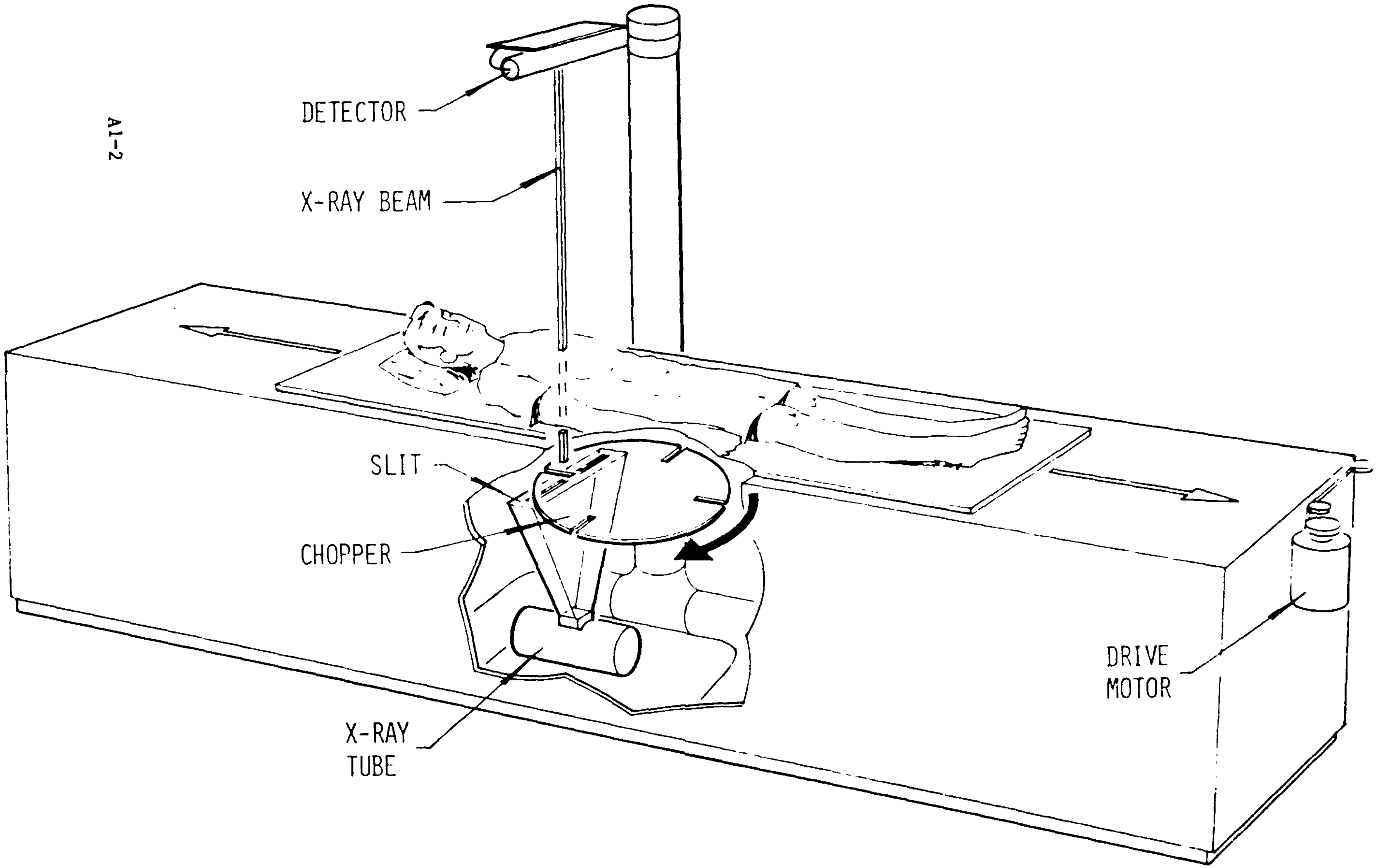

Figure A1-1.   Basic Operation

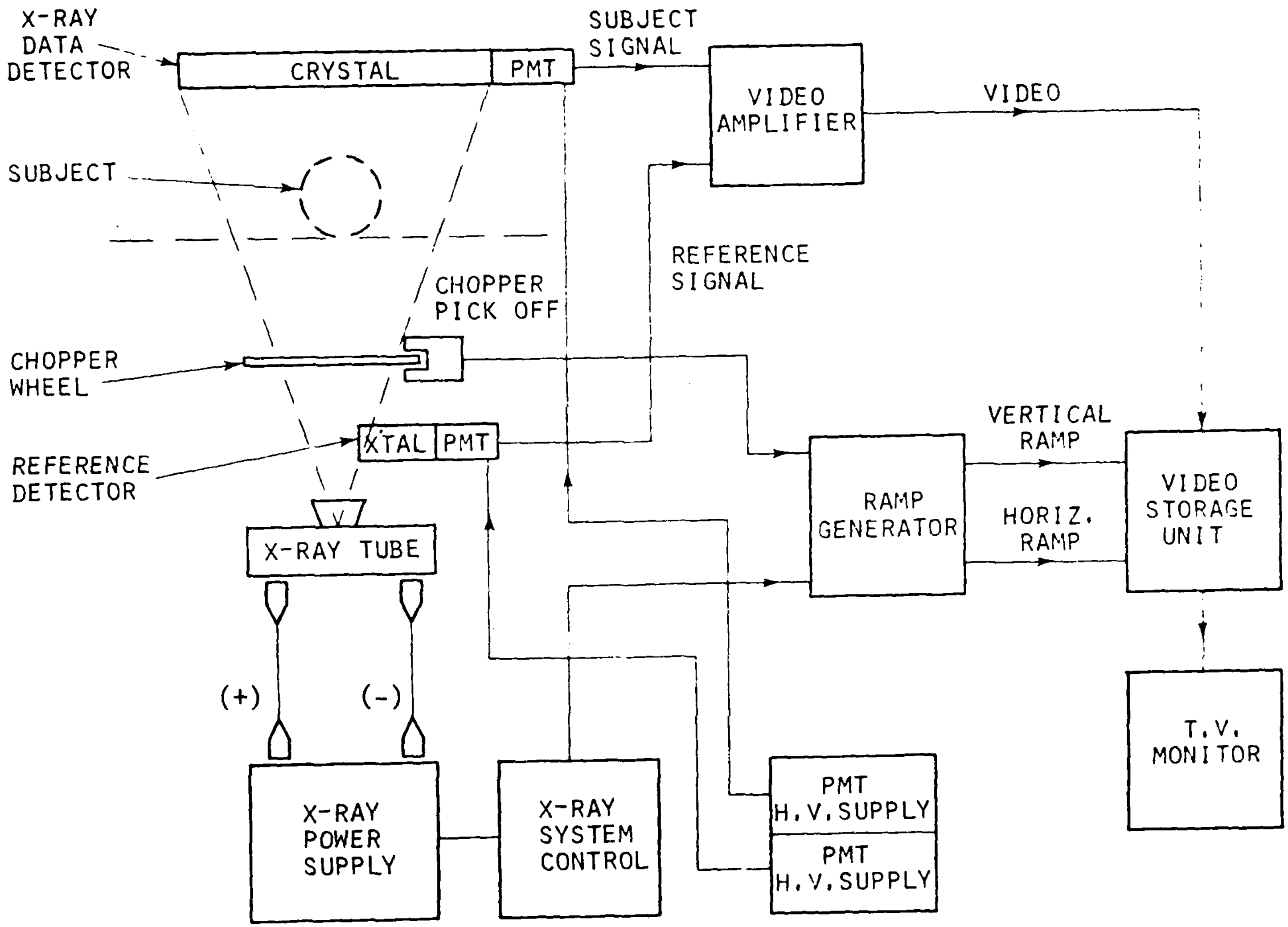

Figure A1-2. Sytem Block Diagram

If the line images of each of these successive scans are displayed one after the other, the result is a complete two-dimensional image of the region under examination. This display is again analogous to the pattern on a video monitor which displays pictorial information as a series of many scan lines. The use of television display techniques is, in fact, well suited to the system since the output of the detector is completely analogous to an ordinary video signal. The same techniques which are used for storing and displaying video signals are used to store and display the radiological images which are generated with the system.

Figures 1-3 and 1-4 are photographs of the entire system with its longitudinal patient transport. The lateral scanning mechanism is housed beneath the table. The unique features of this type of scanning X-ray imaging system are:

a. The high detection efficiency of the X-ray detector results in an exceedingly low radiation dose to the subject examined.

b. The effects of scattered radiation which degrades contrast in a conventional system are eliminated.

c. The image information is stored so that exposure time is limited to that period required to scan the area of interest once (about 10-15 seconds).

d. The image is available for examination instantaneously after completion of the scan.

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f. There is no radiation hazard to the operator from stray radiation, which allows an operator to remain with the patient during the examination.

g. Specially shielded rooms or other facilities are not required.

h. The entire body, or any part of it, may be examined readily.

Figure A1-3. System and Transport

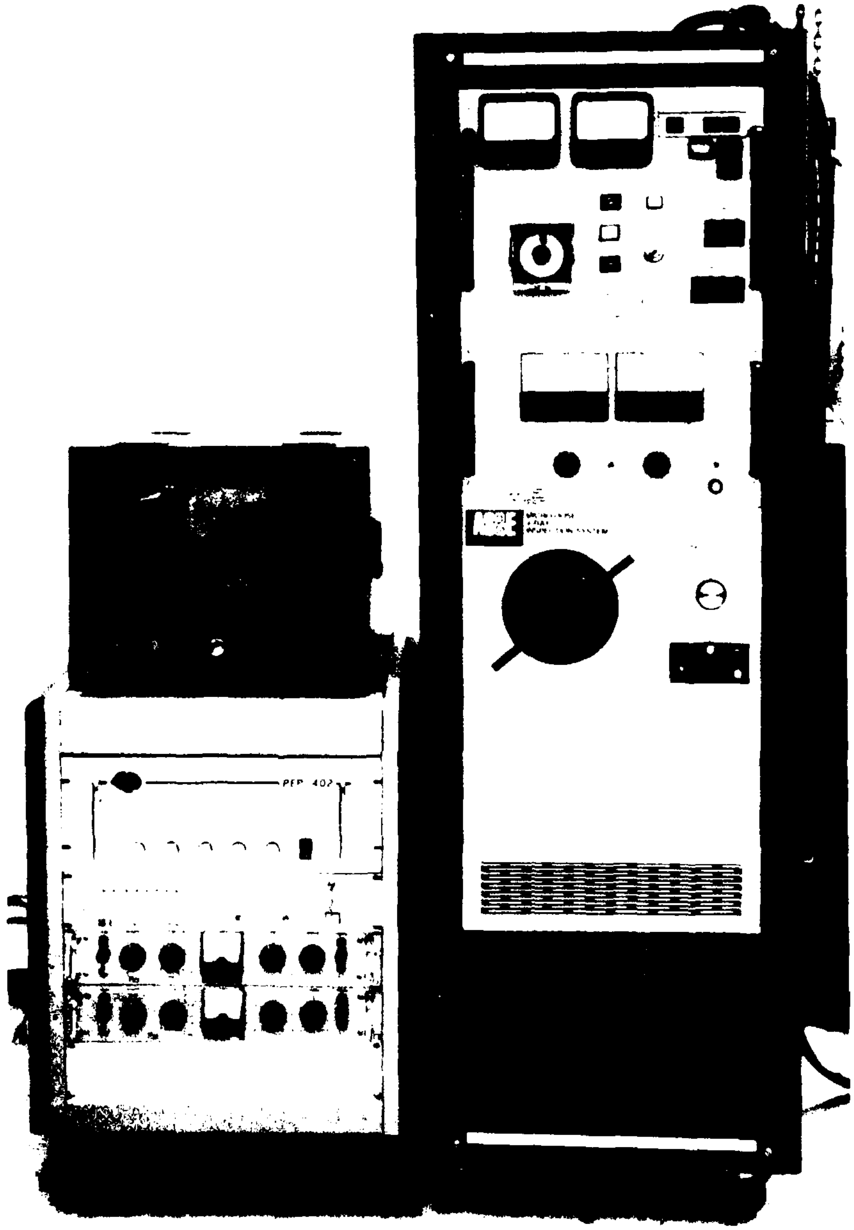

Figure A1-4. Control and Display Consoles

2.0 SYSTEM SPECIFICATIONS

A. HIGH VOLTAGE POWER SUPPLY

1. Input Power: 208 VAC 3 phase, 60Hz @50 Amps
2. Output Voltage: 0-150KV.DC (± 75 KV, referenced to ground)
3. Voltage Ripple: less than 5% peak to peak
4. Output current: 0-100 milliamps
5. Filament Supply Voltage: 0-6 VDC
6. Filament Voltage Ripple: 2% peak to peak
7. Filament Current: 0-6 amps.
8. Output Connectors: Federal Standard

B. X-RAY TUBE

1. Type: Machlett Labs Dynamax HE67B Rotating Anode, Oil cooled
2. Voltage: 150 KV maximum, (± 75 KV)
3. Anode Heat Storage: 300,000 Heat Units
4. Anode Heat Dissipation: 72,000 heat unit/sec.
5. Anode Rotation Rate: 3,300 RPM
6. Target Spot size: 1.0 mm.
7. Target Material: Rhenium Tungsten on molybdenum
8. Filtration: 3mm Aluminum

C. REFERENCE DETECTOR

1. Type: Harshaw Model 852
2. Material: NaI(Tl)

D. SIGNAL DETECTOR

1. Type: Harshaw Model 60 MB 120/1.5-X
2. Material: NaI (Tl)
3. Photomultiplier: Amperex XP-1010
4. Active length: 28"
5. Diameter: 1.5"

E. PREAMPLIFIER

1. Gain: 1-5 (variable)
2. Inputs: Reference signal (V_1), subject signal (V_2)
3. Output: $V_0 = K\ V_2/V_1$; K constant
4. Output impedance: 150 ohms

F. VIDEO STORAGE UNIT

1. Manufacturer: Princeton Electronic Products
2. Model PEP-402
3. Resolution: 1200 TV Lines
4. Grey Levels: 32 linear
5. Video Output format: 1023 line, 30 frame-interlaced

G. MONITOR

1. Manufacturer: Conrac Corp.
2. Model: RQA-14
3. Screen Size: 14" diagonal measure
4. Rates: Automatically synchronized to input signal

H. SUBJECT TRANSPORT:

1. Speed: 0-12 inches per second
2. Weight of subject: 220 pounds max.
3. Drive: Continuously variable DC Motor
4. Length of travel: 72"
5. Transport material: Mylar sheet 0.010" thick

I. MECHANICAL SCANNER - CHOPPER WHEEL

1. Scan Rate: 120 scans per second
2. Scan Angle: 26°
3. Scan Width: 12" at table surface, 28" at detector
4. Scanning spot size: 1mm. square at table surface.
5. Wheel Construction: Aluminum with lead sheet.
6. Chopper slit material: Tungsten

3.0 SYSTEM DESIGN

3.1 X-ray Tube

The conventional X-ray tube consists of an evacuated glass envelope, a heated cathode or filament, and a target or anode. Electrons are generated by the filament of the tube and accelerated toward the anode by a large electrical potential impressed between the cathode and anode. The electrons are decelerated in the anode to produce a beam of X-rays.

X-ray generation is an inefficient process with respect to the large amount of energy necessary to create a useful flux. Most of the electron energy is dissipated in heat which must be taken into account in the design and construction of the X-ray tube. It is also important that the target have as high an atomic number as possible, as the number of X-rays produced is in direct proportion to the atomic number, given constant tube current and voltage. The target must, therefore, have a high atomic number and be able to withstand high temperatures without vaporizing or mechanically distorting. The material that is most commonly used to meet these requirements is tungsten. In the present application, the target is made of rhenium tungsten on a backing of molybdenum. Since the spot size on the target needs to be small (1.0 mm) to achieve good spatial resolution, the energy density at this point can be high enough to heat and vaporize the target material.

The rotating anode tube makes it possible to operate at higher power levels without deleterious effects on the target. Figure 3-1 shows the general construction of the tube. The anode is in the shape of a disc with the cathode positioned off the axis of rotation. Instead of a single spot the electron beam describes a ring on the anode structure. The anode is put into rotation by an external rotating magnetic field before the tube is energized. Each point on the target is only exposed to the full beam for a

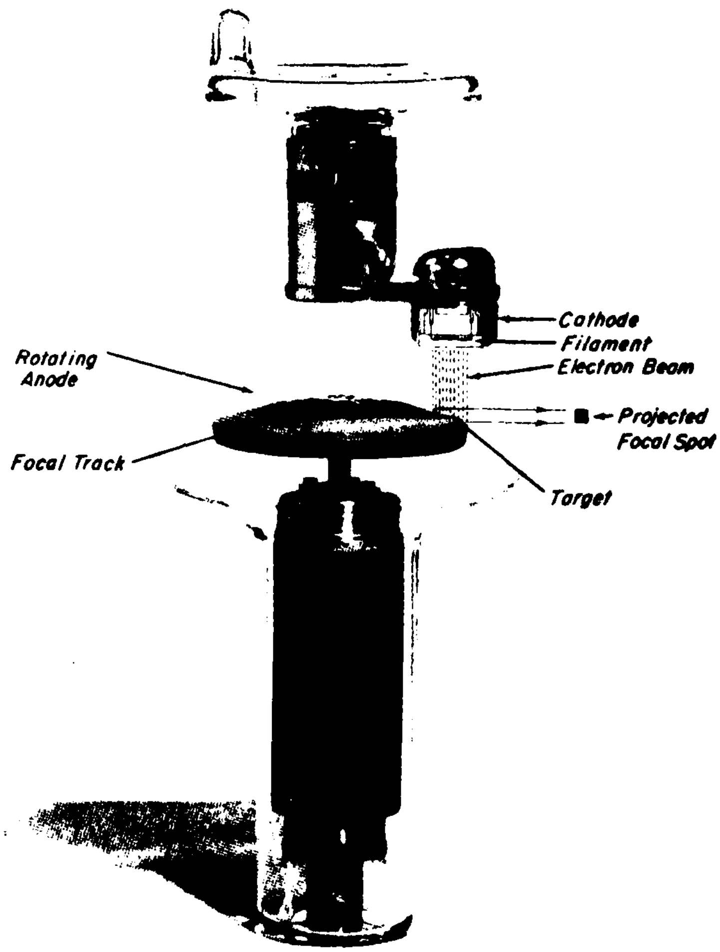

Figure 3-1. Rotating Anode X-ray Tube

fraction of a resolution. In this way the heat load is spread over a large area of the target.

The amount of energy stored in the target following an exposure is the product of tube voltage, current and the exposure period, and is expressed in "heat units." Safe operation of the tube requires that the heat stored as a result of an exposure be dissipated before the next exposure is begun so that the total heat stored in the target at any particular time does not exceed the target capacity. For example, if the tube is to be operated at 150KV and 100 ma, the rate of heat accumulation would be:

$$150\text{KV} \times 100 \text{ ma.} \times 1.35 = 20,\ 250 \text{ H.U./sec.}$$

(Note: The constant 1.35 is applied where the power supply is 3 phase which is the case here.)

This would allow a maximum operating time of about:

$$300,000/20,250 = 14.8 \text{ seconds}$$

The target cools at an average rate of 72,000 H.U./min. Therefore, the tube would need 4 minutes to cool down before another 14 second exposure is begun. Figure 3-2 shows the cooling chart for the Dynamax 67.

The normal operating parameters for the present system are 110KV, 50 ma and 15 seconds. At this level, the heat stored in the target per exposure is 111,000 H.U. If a series of exposures is to be taken, a proper duty cycle must be observed so that the aggregate heat load does not exceed the 300,000 H.U. In order to balance a heating rate of 111,000 H.U./exposure, a waiting period of 1.55 minutes must be observed between exposures.

ANODE THERMAL CHARACTERISTICS

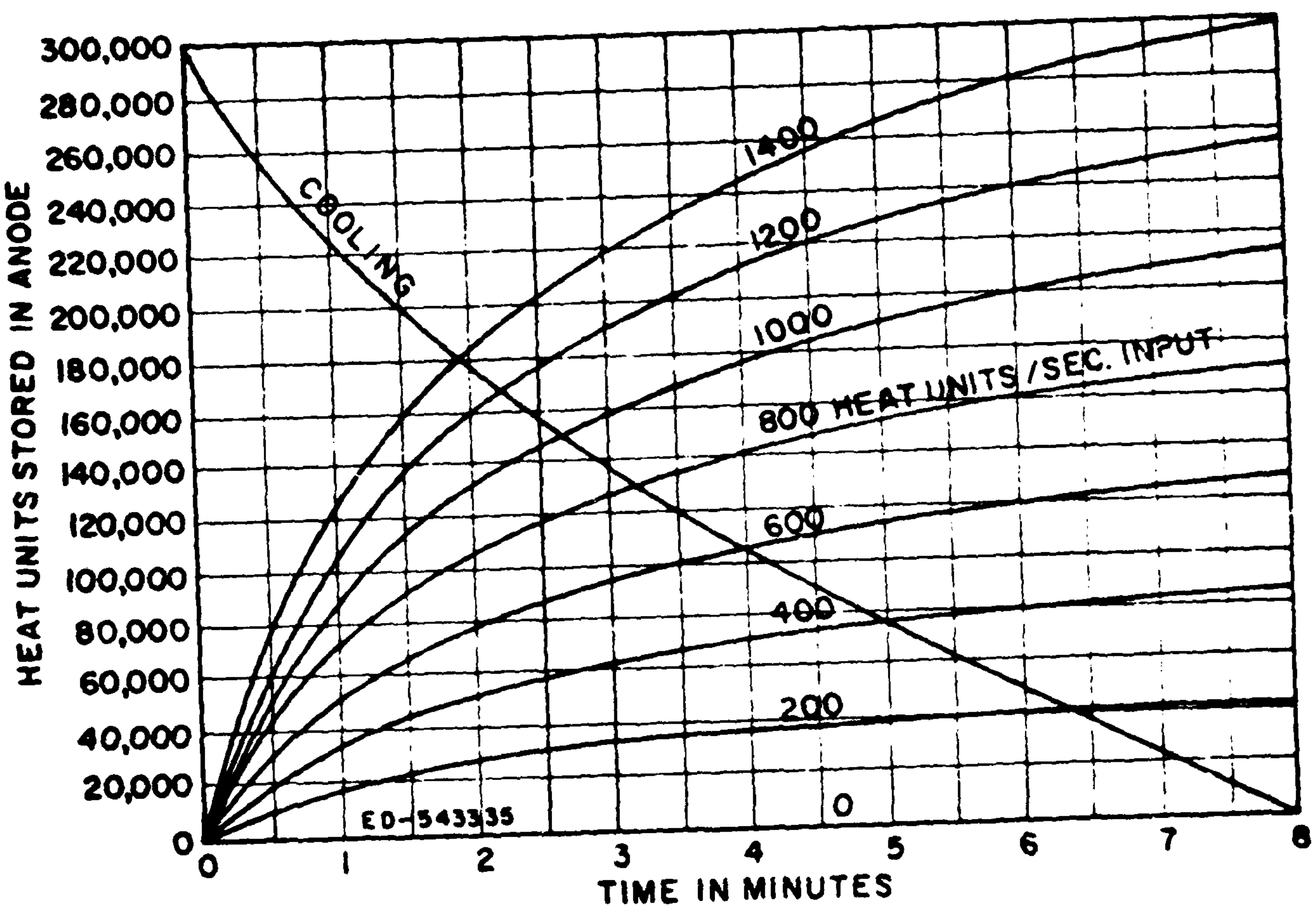

Figure A3-2. Cooling Chart

The X-ray tube is operated with an integral system for rotating the anode and for cooling the tube. The rotor controller provides power to the rotor coils and performs other control functions for the system. It is necessary that the X-ray tube be operated with a proper sequence of events which is programmed by the controller.

The filament must be held in a standby mode for some period prior to operation of the tube at high power. The standby mode energizes the filament to low power level and allows the filament and support structure to reach a temperature where the filament is ductile.

When the tube is operated, a command is initiated at the input of the motor controller. The controller in turn, accelerates the rotor to operating speed, boosts the filament from a standby to an operational current level, and turns on the high voltage after a two second time delay. Once the exposure period has been completed, the controller turns off the high voltage and returns the tube to the standby mode, with the power removed from the rotor.

An oil cooler is connected to the tube with air-tight connectors and is permanently wired to the X-ray tube power supply. Whenever the power supply is turned on, the cooler automatically comes on. Since the temperature will rise rapidly inside the housing if the circulation of the oil is stopped or restricted, the tube housing has a temperature sensitive switch inside which removes power if the housing temperature exceeds a safe value.

The bearings of the rotating anode tube must operate inside the vacuum of the tube envelope and are subject to wear. It is possible to operate the tube continuously with rotor stationary, providing that the power level is low, which is advantageous for alignment and test procedures. Precaution must be taken with this

type of operation because, with the rotor stopped, excessive power can damage the tube. The maximum continuous input power level to the tube in this mode of operation is 1,250 heat units per second.

3.2 High Voltage Power Supply

The high voltage power supply provides all the necessary power to the X-ray tube. The supply has some unique requirements owing to the special demands of the X-ray tube. Figure 3-3 is a circuit diagram of the X-ray tube and power supply. The supply should have little voltage ripple without using capacitive filtering at the output because the scanning X-ray technique is ripple sensitive. The presence of capacitance in the voltage output allows for the storage of charge which could be damaging to the tube should a momentary arc occur.

The method of obtaining high voltage with low ripple is by the use of a "12 pulse-three phase" transformer. A single phase rectified voltage waveform has 100% ripple. Rectifying a three phase waveform reduces the ripple to 13%. A further variation on the three phase technique is the "12 pulse" system whereby the secondary of the high voltage transformer is connected in a "Y-delta" manner. The two different secondary connections introduce a phase shift between the two sections of the supply, which further reduces the ripple to 3.5%. The resultant ripple has twice the frequency of the conventional three phase supply. The term "12-pulse" applies to the fact that within one cycle of the 60. Hz input waveform the rectified voltage peaks or ripples 12 times.

The filament transformer is insulated primary to secondary by at least half the total tube voltage. The filament of the X-ray tube nominally requires 4 to 6 volts at 3 to 5 amps. The filament in this system is powered by direct current in order to further suppress ripple in the tube current. It was found experimentally that if the filament is powered by 60 Hz. power, the thermal time

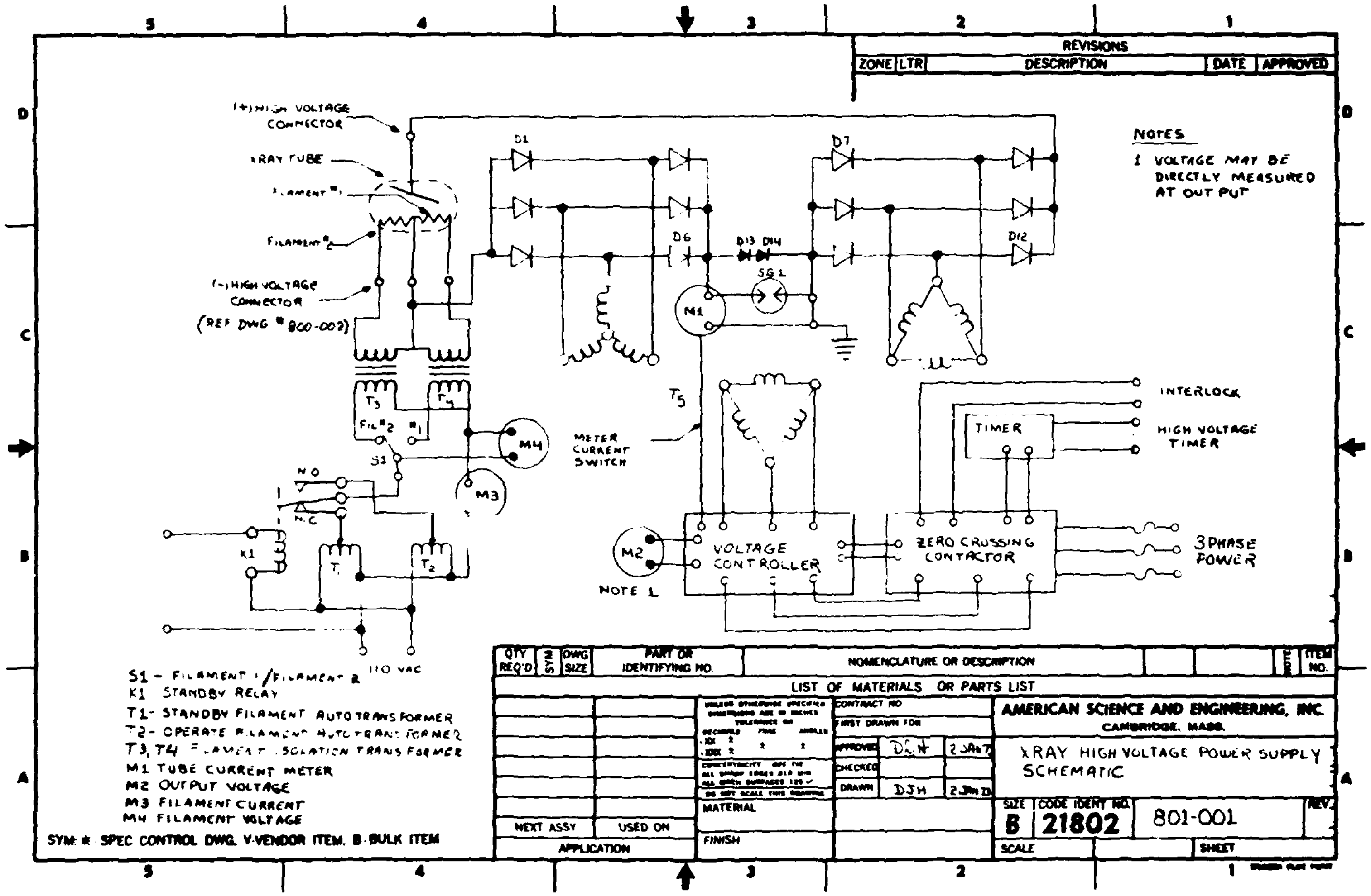

Figure 3-3. X-ray High Voltage Power Supply Schematic

constant of the filament is not long enough to filter variations in filament emission with the supply current.

Physically the filament power supply is located inside the high voltage tank next to the output connector. A filament reversing switch is incorporated as part of the controls. This switch controls a reversing relay which, upon command, will reverse the current through the filament a procedure which serves to lengthen the life of the filament.

The power supply has two output connectors, one positive and one negative. The negative connector also has the connections for the filament as integral parts. The high voltage cable has separate conductors for this purpose. Figure 3-4 shows the connector and the wiring of the tube. The power supply tank construction is basically transformer and rectifier stacks mounted on a nonconducting framework and immersed in transformer oil. The tank is shown in Figure 3-5. Voltage and current are read directly and these signals are routed to the panel meter through the control cable. Voltage is monitored by means of a high resistance divider inside the tank and the current is directly measured at the junction of the positive and negative sections of the supply. Both signals are low voltage with respect to ground and each signal wire is protected by a spark gap to prevent high voltage surges from reaching any part of the system accessible to an operator. The tank is interlocked so that the cover cannot be removed without disabling the power supply. This is done with switches which are located at the bottom of the high voltage tank. Figure 3-6 is the schematic for the power supply.

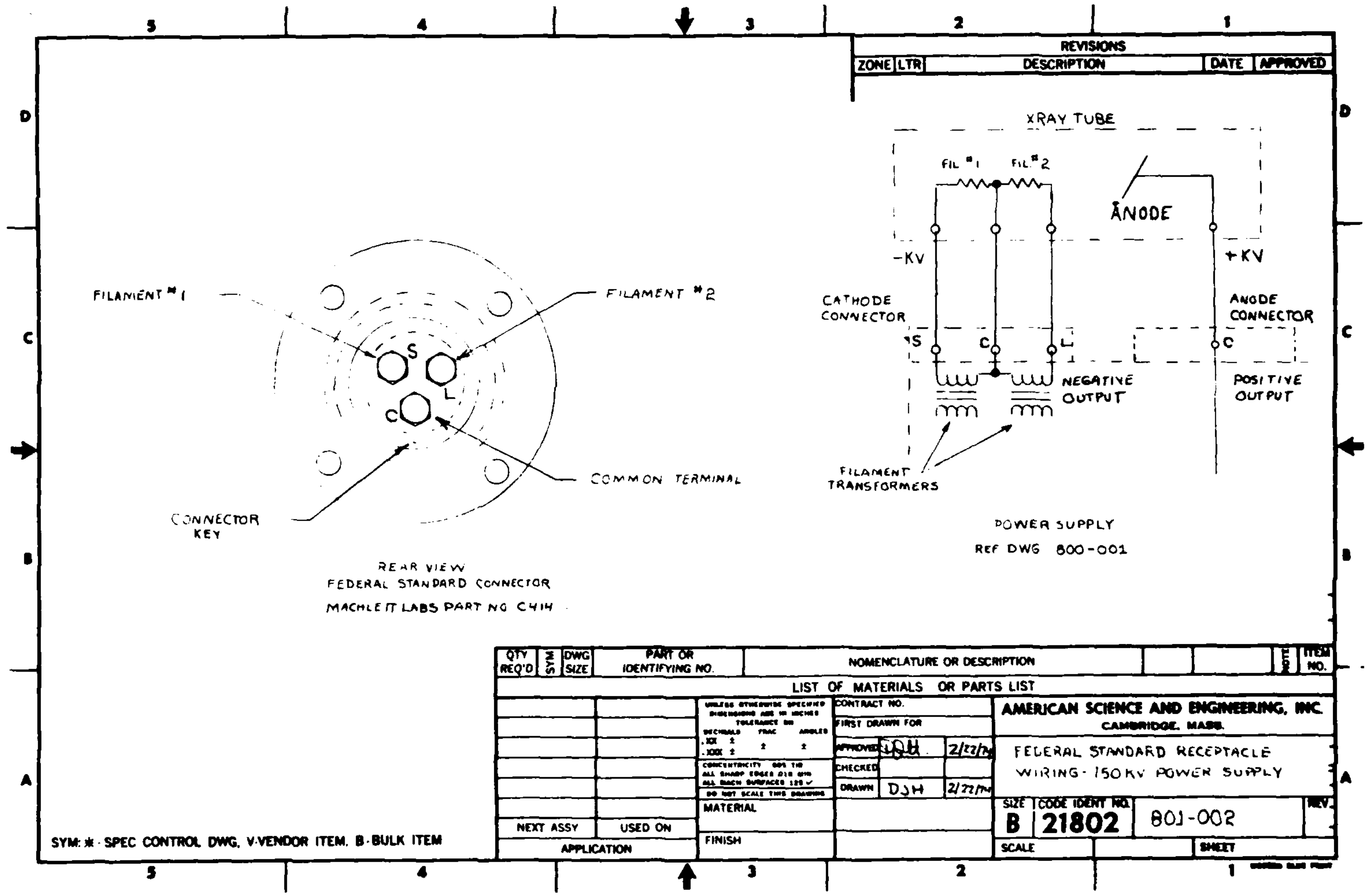

Figure A3-4. Federal Standard Receptacle Wiring - 150KV Power Supply

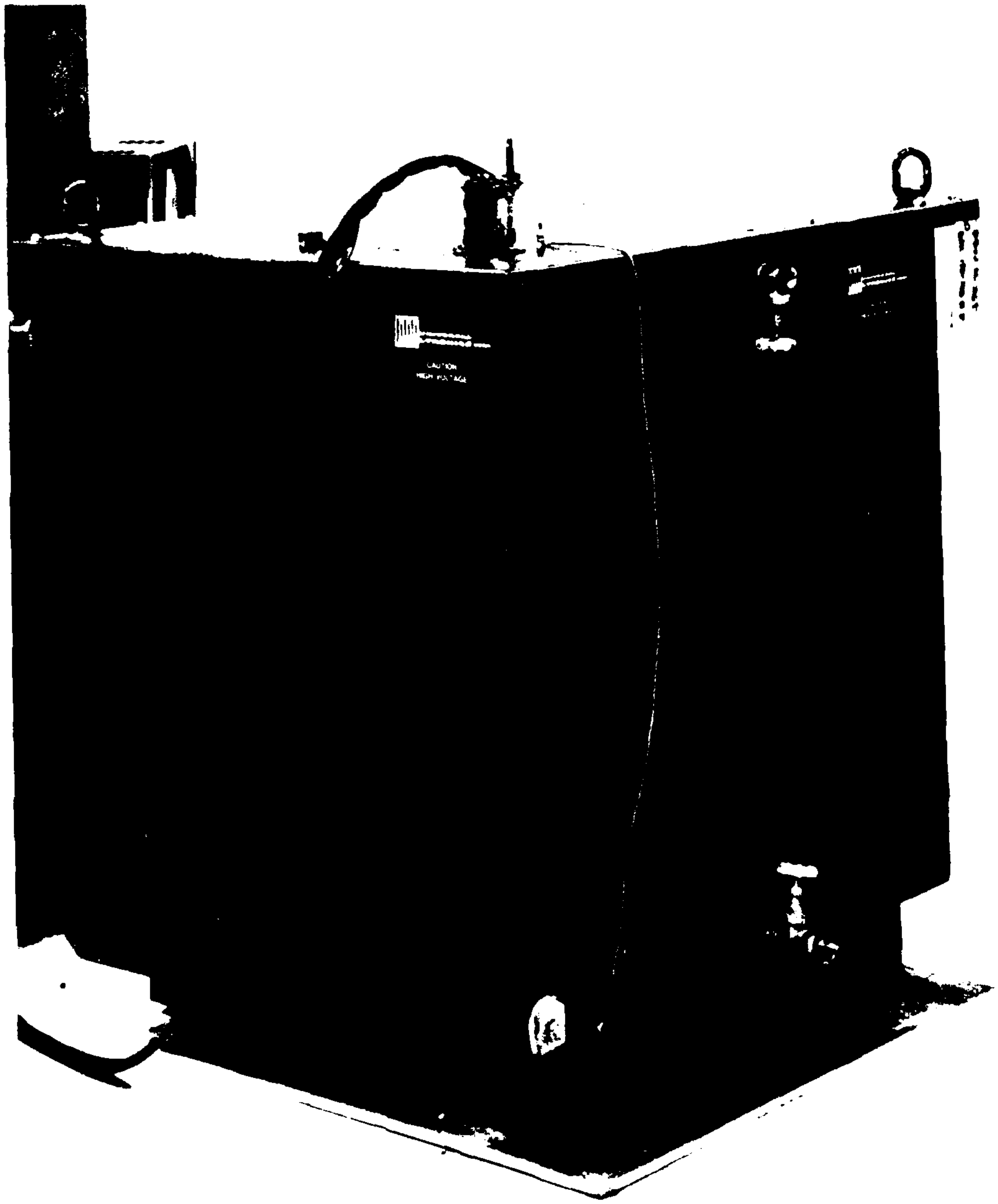

Figure A3-5. High Voltage Tank

The entire high voltage framework lifts out of the oil tank for service and maintenance. The nonconducting framework is suspended from the top cover of the supply so that lifting the top cover three feet exposes all components. The tank is filled through ports in the top. The oil used for insulation is transformer grade oil such as Shell Diallyl AX or the equivalent. The power supply tank holds 187 gallons of oil. The oil contains inhibitors to prevent carbonization due to corona and arcing. It is necessary that the oil be free from contaminants and dirt.

3.3 Power Supply Controller

The high voltage power supply tank is powered and controlled by this unit. The electrical schematic is shown in Figure 3-7 and the front panel in Figures 3-8 and 3-9. The controller consists of five main sections: 1. Circuit breakers, and controls, 2. High voltage autotransformer, 3. Solid state contractor, 4. Filament controller and 5. X-ray tube controller.

The high voltage is controlled by the autotransformer which varies the AC voltage into the tank from 0 to 208 volts AC, which corresponds roughly to 0-150 KV at the output to the X-ray tube. In addition, the filament controller provides an adjustable 0-8 VAC to the filament transformer. The sequence of operation of the controller is determined by the requirements of the X-ray tube. The main circuit breaker controls all power into the unit. When the X-ray exposure sequence is begun, the rotor boosts the filament, accelerates the rotor, and after two seconds, turns on the high voltage. The unit which actually turns the high voltage on is the solid state contactor shown in Figure 3-8 which consists of SCR's, rectifiers and control logic. This unit turns each phase on as it passes through the zero voltage point of its waveform. The reason that zero crossing contacting is important is that it provided a smooth voltage and current start-up rather than a large surge as would be produced if the power was turned on

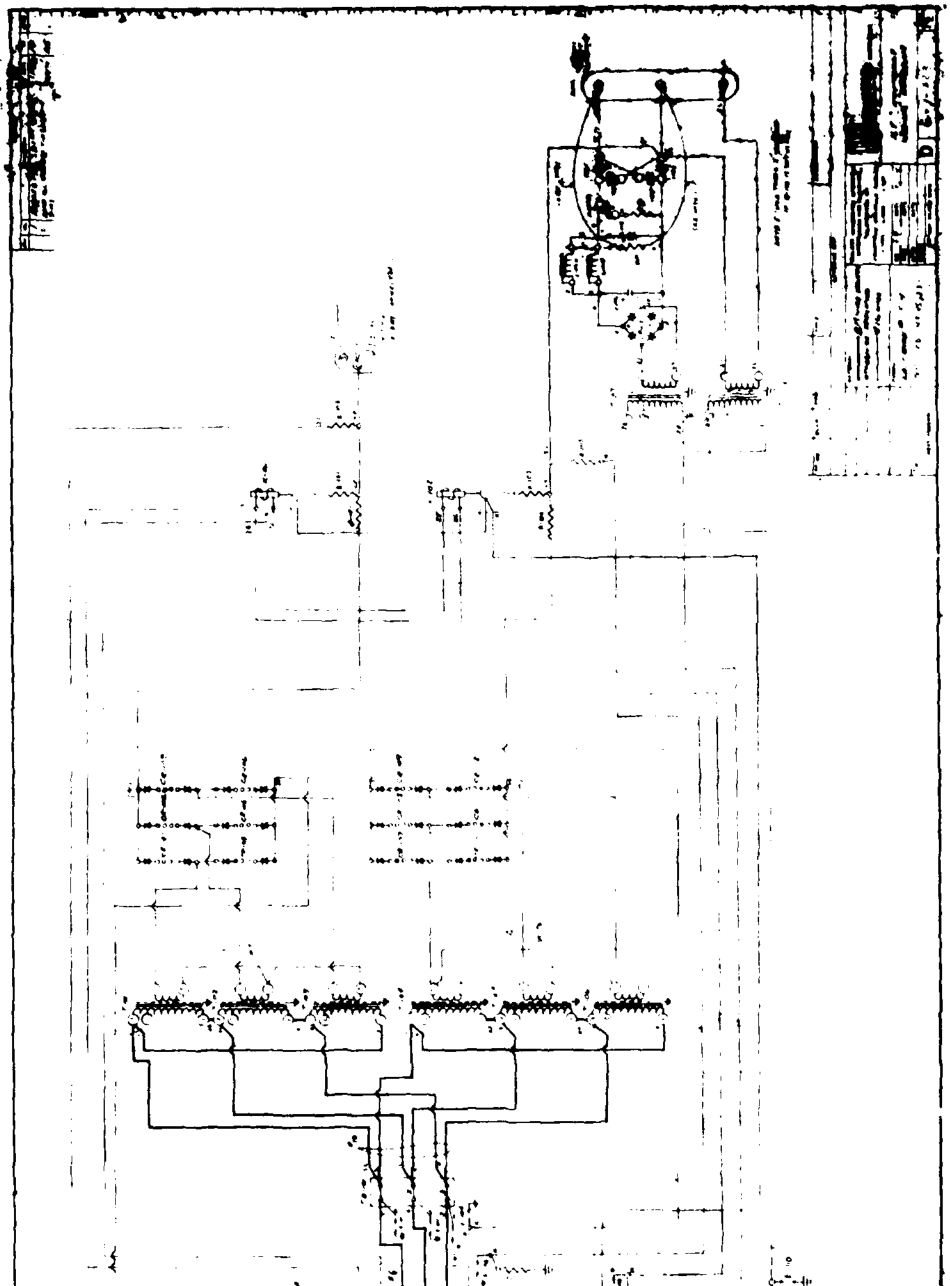

Figure A3-6. HV Assembly Wiring Diagram

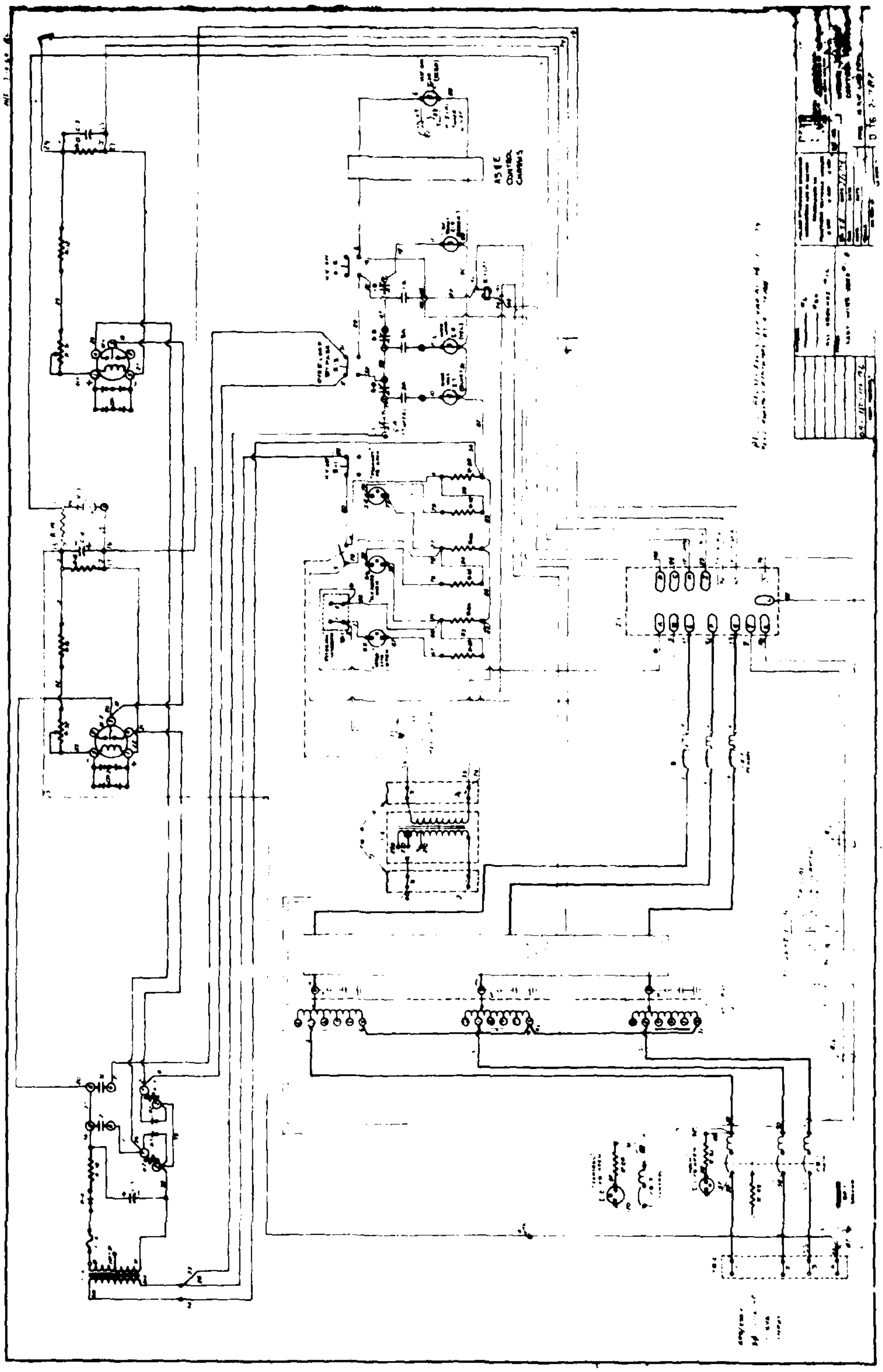

Figure A3-7. Wiring Diagram Control Assembly

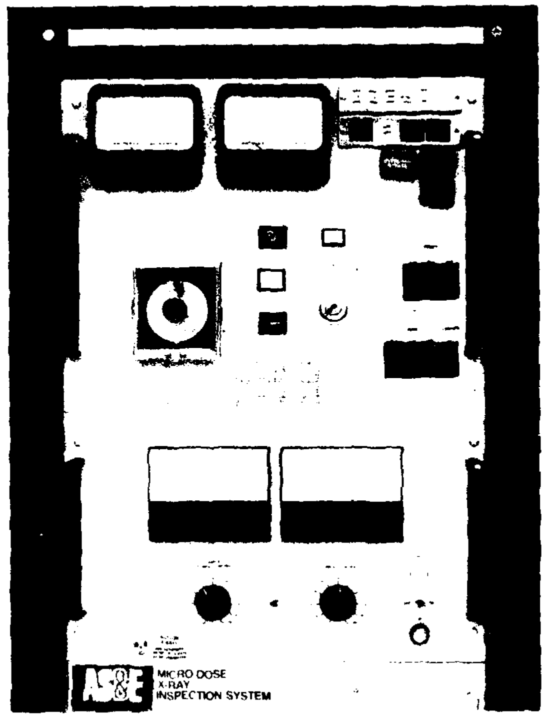

Figure A3-8. Power Supply Controller Front Panel

Figure A3-9. Power Supply Controller Front Panel

at an arbitrary point with a simple switch. This is necessary with a large power supply, since heavy inductive surges can damage components.

The solid state contactor brings the supply to full output in under 8 milliseconds and produces virtually no electrical interference. Once the high voltage comes on, it remains on until the timer, which is on the front panel, completes its timing cycle. At the end of the cycle, the high voltage power supply shuts off and the tube returns to the standby state.

The power supply has several features to protect the X-ray tube and the supply. The supply has two sets of circuit breakers which prevent the supply current from being run at more than 100% of rated power. If the output current is short circuited, the supply is only capable of an instantaneous current of 110% of full load current. The circuit breakers will open the circuit should this occur. In addition, the direct reading current meter has a cursor to set that level of current at which the supply will automatically be turned off. This feature is primarily intended to protect the X-ray tube, but also serves as an over-current protector. The X-ray tube thermal overload switch is also wired into this meter circuit to prevent the supply from being energized when the tube housing is too warm.

The high voltage meter also has a set point needle, which prevents the supply from being run beyond a set voltage. Normally, this needle and the over-current needle are set just beyond the operating points of the tube so that the reaction time is minimal.

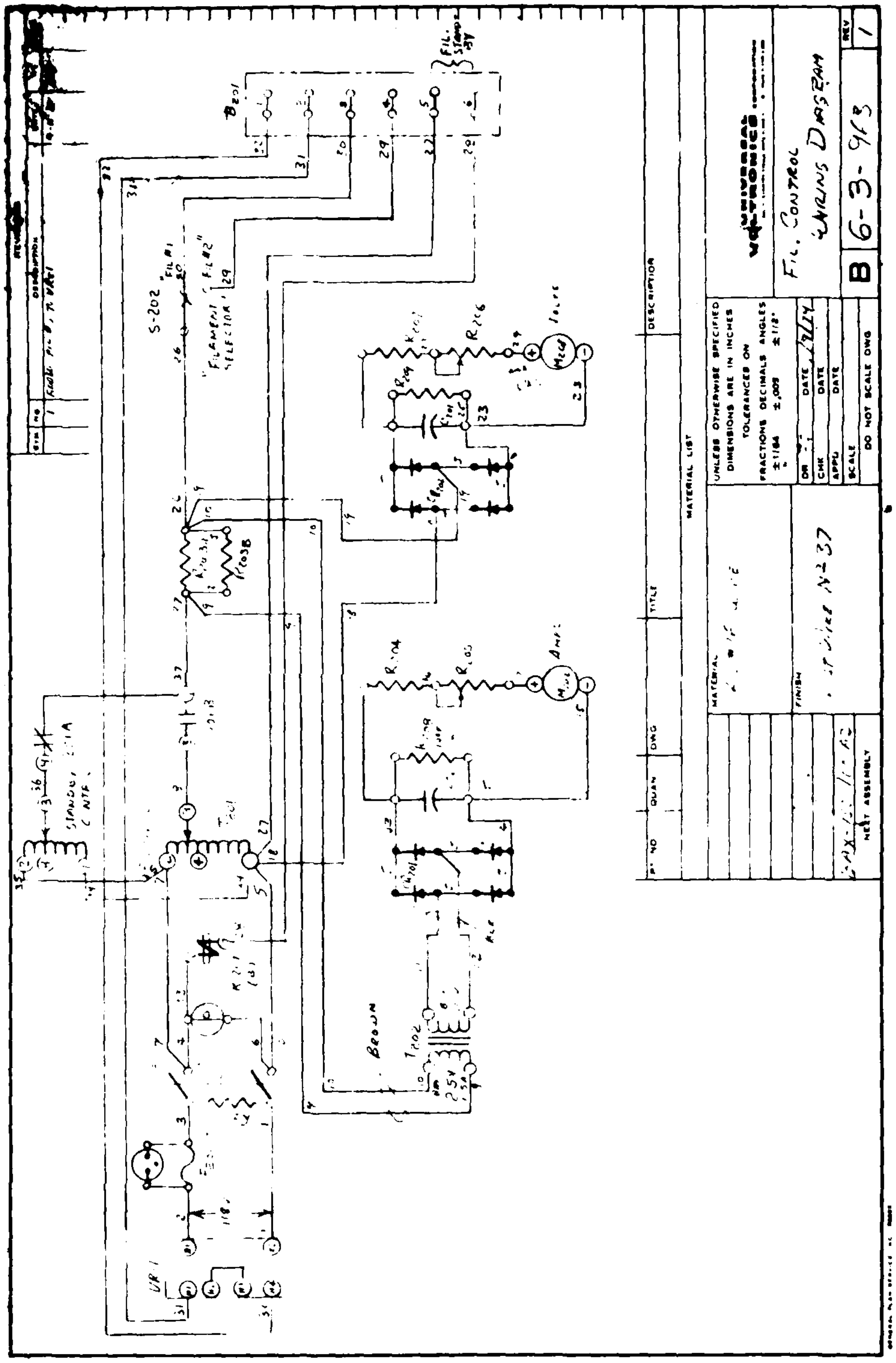

Figure A3-10. Filament Control Wiring Diagram

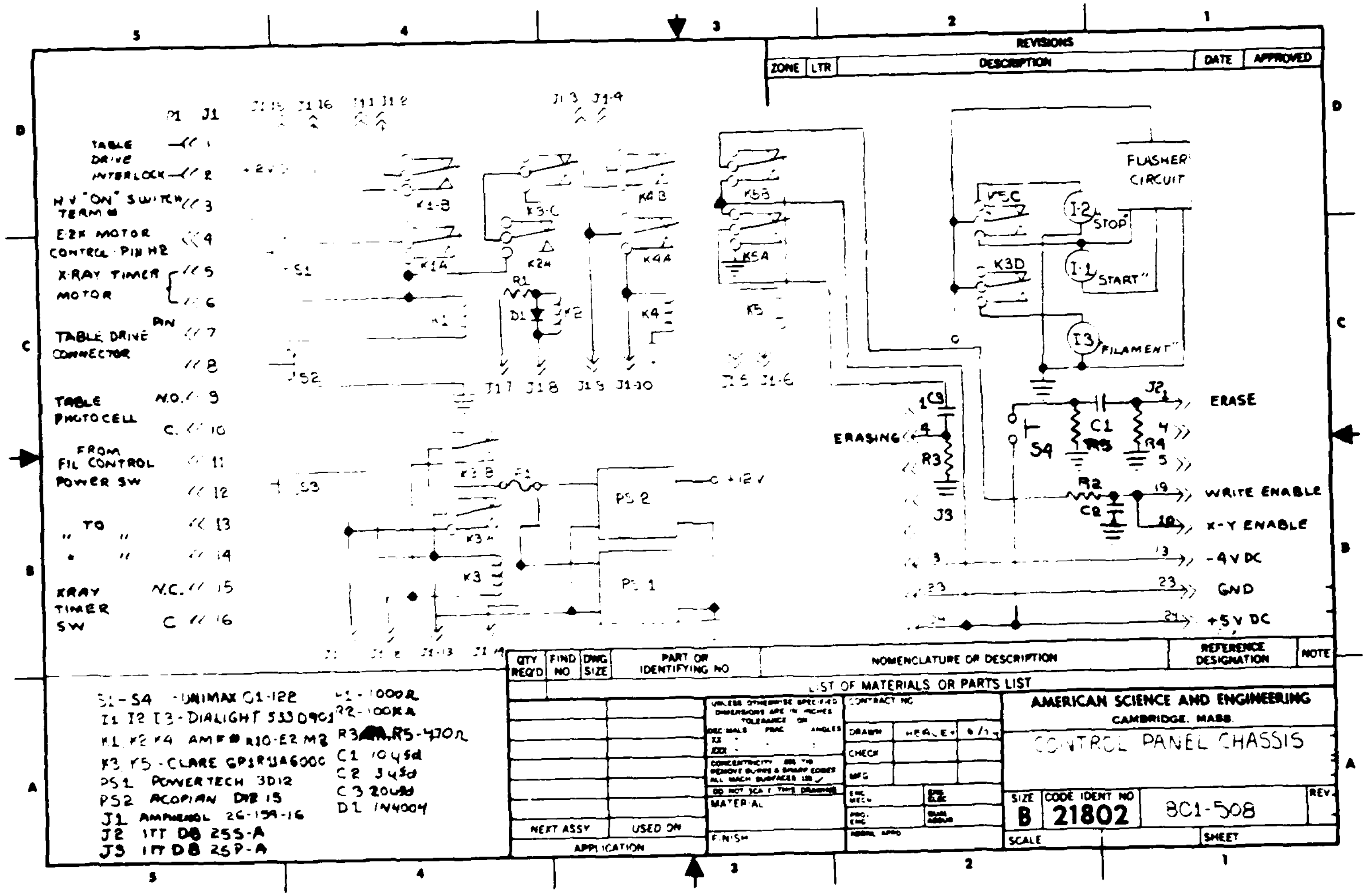

Figure A3-11. Control Panel Chassis

The filament control unit has two modes of operation, standby and operate. In the standby mode, the tube is powered to keep the filament and its support structure warm enough to prevent thermal shock and instability when the tube is energized. When the supply is switched to the operating sequence, the filament controller automatically switches to a second transformer setting which powers the filament to the operating temperature, which corresponds to the desired X-ray tube current. The filament unit automatically returns to the standby mode when the power supply is turned off. Figure 3-10 is the schematic of the filament control.

3.4 Control Chassis

The control chassis serves to interconnect the power supply and the rest of the X-ray system and is mechanically attached to the power supply control panel. Figure 3-11 is the schematic for the control chassis. Push buttons for controlling the entire system (S1, 2, 3) are on the front panel of the power supply panel. S3 is the filament power switch which latches the filament supply relay when pressed. This prevents the filament from being unnecessarily powered when the breaker is turned on. The filament switch also energizes the logic and control power supplies (PS 1 & PS 2) which power the control logic and ramp generator. The X-ray/Table start button S1 has two functions; it allows the table drive to run and it initiates the cycle for turning on the X-ray tube power supply.

Depressing S1 latches relay K1 if the X-ray timer switch is closed. The timer switch closes only when the power supply is on. The Table start button is used primarily to position the table and later to start the system. Relay K1 supplies control power to K2 whose coil is connected across the armature of the table drive motor. Unless the motor is running in the proper direction and at a minimally safe speed, the control voltage stops at this point. Resistor R1 and Diode D1 control the speed and

direction parameters. R1 is set so that the relay will not close unless the motor is running at at least the voltage which corresponds to 0.8 inches per second. D1 shorts the relay if the motor direction, i.e., the motor polarity, is improper. When K3 closes, the control voltage is routed to K4. K4 is latched when a closure is received at pins 9 and 10 of J1. These pins are connected to a photo-detector mounted on the X-ray detector mast. The detector has a light source and a photocell. The light source is focused on the edge of the subject transport table. When a special reflector is laid on the table on the edge of the mylar, it will trigger the detector when it passes this point. The light reflected back to the photocell triggers a relay which starts the system. Because of the nature of the reflector and the detector stray, reflections will not energize the system and X-rays will not be produced unless the reflector is in place and passes under the photocell. Once the closure occurs, the entire system is latched on and the X-ray power supply and table run for the duration of the time set on the timer; however, should the table stop or run to its full travel, K1 will unlatch and shut down the system. Likewise, the timer running to its term will unlatch the relay. The only time that the full system will latch for a complete X-ray run is when all conditions necessary for safety of the subject and proper operation of the X-ray tube are met. Each condition is monitored and fed back to the control chassis to interrupt operation should any abnormal occurrence take place. In addition, it is necessary that the X-ray start button be depressed for at least the time period required to complete the X-ray tube sequence (2-3 seconds) or the system will not latch. This prevents the supply from being activated by an accidental momentary pressing of the button. Pins 3 and 4 of J1 provide the closure to the rotor controller, which in turn energizes the power supply.

Relay K5 provides control signals to the system logic and to the video storage terminal. When the high voltage is turned on and the timer is energized, K5 closes. This provides a command to the ramp generator to begin sweeping and generating the video blanking for proper rcording. Also, a command sets the video storage unit into the writing mode. When the cycle is complete and K5 opens, the video storage unit returns to a READING mode and the ramp generator resets. Normally the erase of the video storage unit is accomplished by pressing the ERASE button on the front panel. It is also possible for the erase to be automatic by deriving the erase command for the normally-open contact of K4A.

The lights I-1,2, and 3 are in the pushbutton sockets associated with each of the functions. When the relay controlling a particular function closes or opens, the auxiliary contacts on that relay provide power to the indicator. In addition, there is a flasher circuit on the stop button. The function of the flasher is to warn the operator of the duty cycle necessary for full power operation of the X-ray tube. If the tube is energized for any period, the flasher will continue to turn the stop light on and off for a period of 2 minutes following the exposure. The two minutes is a safe period for repetitive operaton of the tube at 110KV and 50 MA.

The control chassis has three connectors which are wired into the power supply control harness. The control buttons are permanently wired to the chassis to prevent loss of control.

3.5 The X-ray Chopper Assembly

The X-ray tube is physically mounted to the chopper assembly as shown in Figure 3-12. A 30° wide beam of X-rays projects upward inside the lead cone to the chopper wheel. The longitudinal slit shown in Figure 3-13 narrows the beam to a fan 1 mm. wide at that point. The chopper wheel is a lead filled aluminum shell with

four sets of tungsten slits mounted in quadrature, as shown in Figure 3-14. When the wheel rotates, each of the slits, in turn passes over the entire length of the longitudinal slit. The projection of the fan beam on the slits of the rotating wheel results in a narrow scanning beam of X-rays which repeatedly moves in one direction over the 30 degree spread of the fan beam. The spacing of the slits is 1 mm wide so that the beam dimension at the chopper wheel or just above it is 1mm x 1mm.

The chopper wheel is driven by a synchronous motor which rotates at 1800 rm so that the slits pass at a rate of 120 scans per second. The wheel is mounted on the shaft of the motor.

The chopper wheel must be synchronized to the scanning electronics to allow for proper registration of the image in the storage unit. Therefore, it is necessary to provide an accurate signal at the instant the sweep of any particular slit begins. This is accomplished by a chopper wheel pick off. The pick off, shown in Figures 3-15 and 3-16 consists of an incandescent lamp on a bracket and a silicon phototransistor and lens assembly. The detector is physically located at a point that corresponds in time to approximately 0.5 msec prior to the beginning of the scan. The lamp illuminates the slits as they pass, and when the detector senses the passing image of the slit, a signal is given.

This signal is processed by the chopper signal amplifier to provide an electrically clean trigger with minimal jitter. The schematic of the chopper amplifier is shown in Figure 3-17. The signal is sensed by the LM 311 comparator. When the signal exceeds the voltage determined by the ratios of $\frac{R_3}{R_3 + R_4}$, the comparator output turns Q_2, the 2N2222 transistor on. Q_3 is an inverting stage for compatibility with the ramp generator. The photodiode PD1 is used to align the system physically. It turns off whenever the cell is illuminated and is used to position the

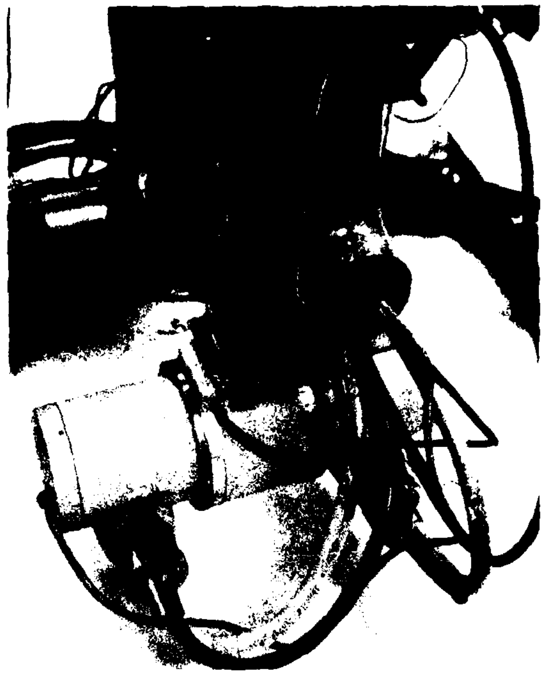

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Figure A3-12. X-Ray Tube Mounted to Chopper Assembly

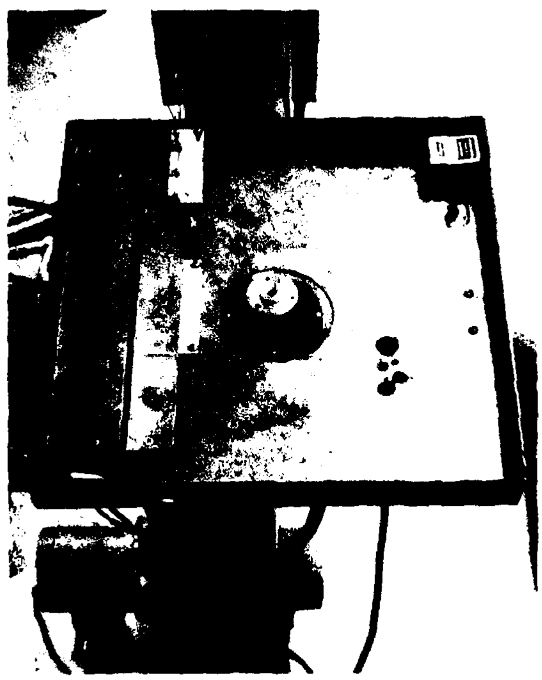

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Figure A3-13. Longitudinal Slit

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Figure 3-14. Chopper Wheel

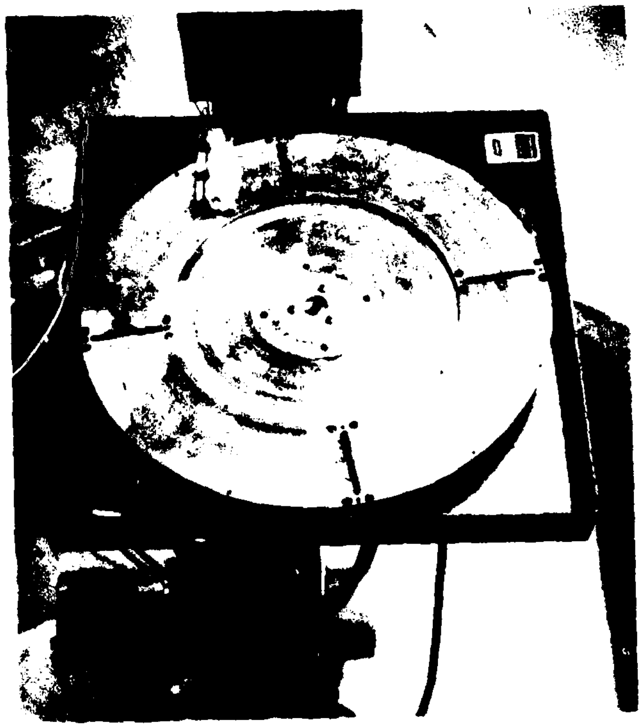

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Figure A3-15. Chopper Pick Off

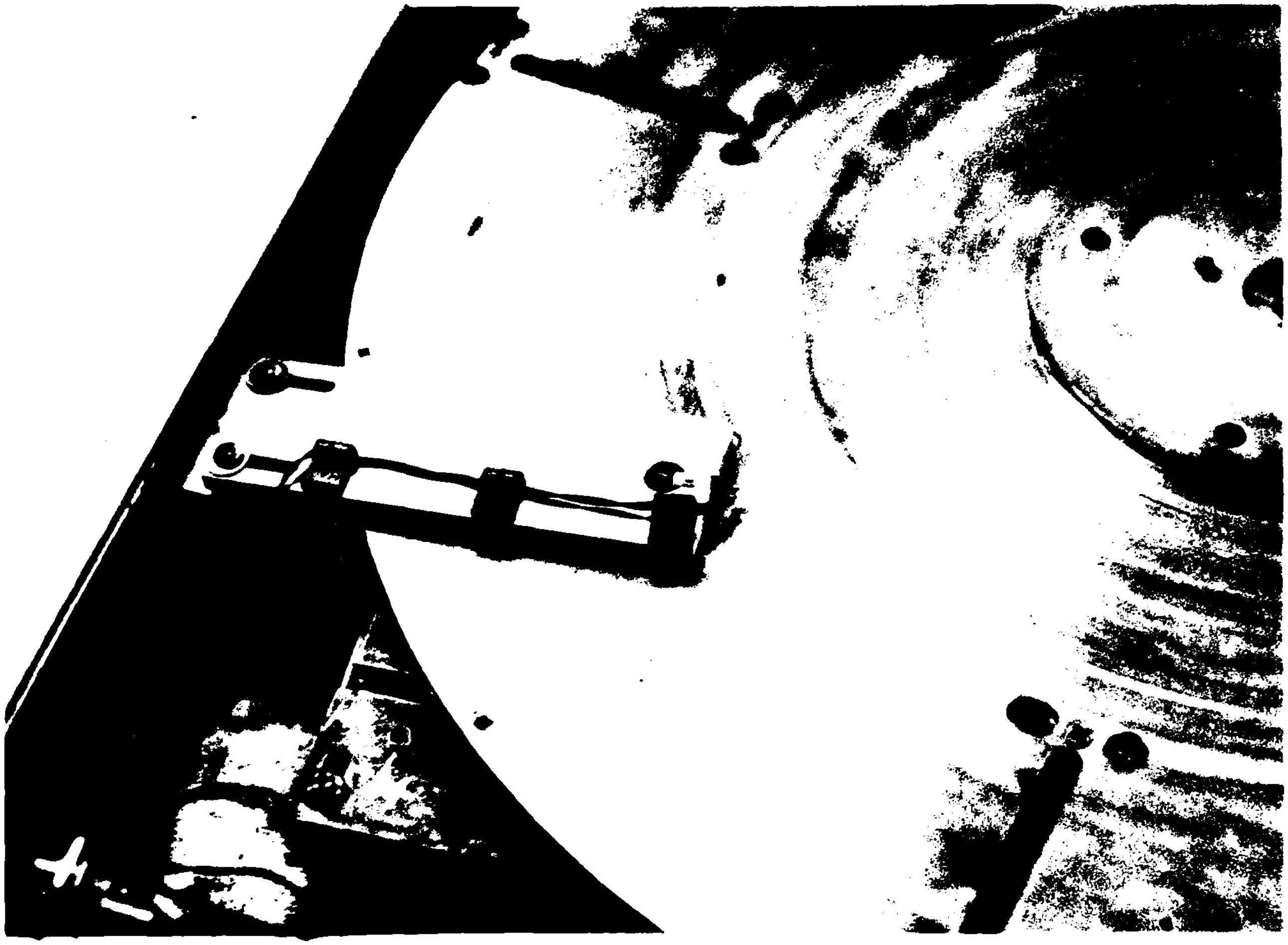

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Figure A3-16. Chopper Pick Off Lamp

light source when the wheel is in place. The gain adjustment, R_2, is set to trigger the comparator at the point of the light pulse that provides the least jitter. The adjustment is made by observing the light pulse and triggering a scope with the output of the comparator. The superimposed light pulses will fluctuate the least when the jitter is minimum.

3.6 Detectors

The X-ray detectors used in this system consist of NaI(Tl) inside a ligth-tight case with a photomultiplier viewing the interior. When X-rays strike the crystal of sodium iodide, it scintillates or produces light impulses. The number of scintillations is proportional to the number of X-ray photons striking the crystal and the brightness of each scintillation is proportional to the energy of the X-ray. The photomultiplier responds to these pulses and amplifies them with a noiseless gain in the region of 10^6.

The detection scheme used with this system utilizes two detectors, one as a reference detector and one as the data or subject detector. The reference detector has only a small disc of sodium iodide two inches in diameter and one inch thick. The data detector has a column of sodium iodide, 1.5 inches in diameter and 28 inches long. This detector needs to be long in order to detect the scanning X-ray beam over the entire 28 inches of scan dimension. The physical locations of these detectors is shown in Figure 3-18.

Ripple on the high voltage power supply for the X-ray tube manifests itself at the output of the data detector with a much higher percentage of the signal voltage than that of the voltage on the X-ray tube. This is due to the fact that the total energy of the X-rays emitted is proportional to the square of the X-ray tube voltage. Therefore, at 100KV, if the power supply has a ripple factor of 5% the ratio of X-ray energies at the high and

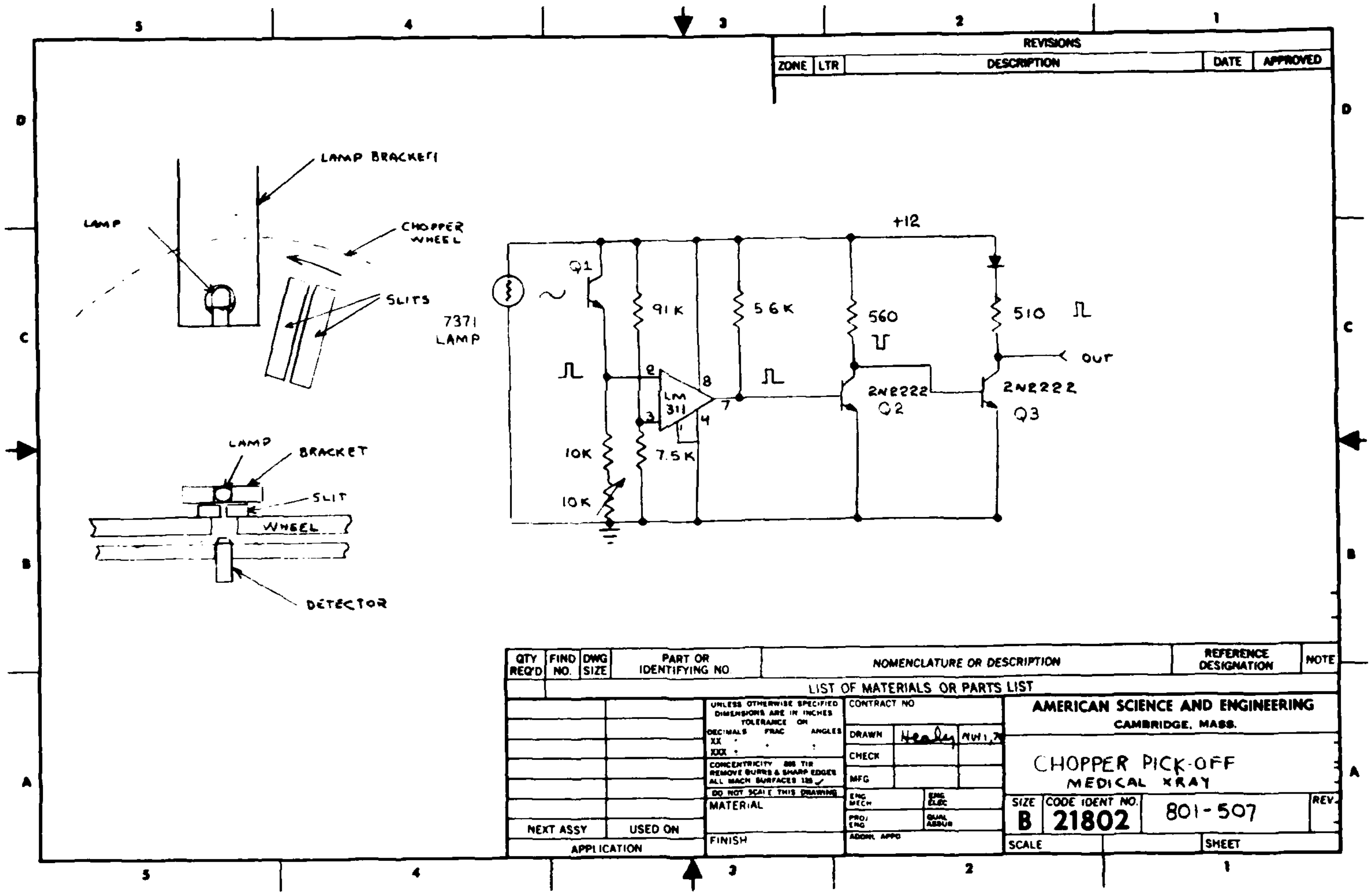

Figure A3-17. Chopper Pick Off Medical X-ray

low points of the waveform is about 10%. It was noticed experimentally that the signal ripple factor indicated a higher third power dependence on the voltage ripple. The ripple under this circumstance is about 15%. The consequence of this fact is that the ripple appears as an unwanted signal, which causes variations in the image on the video display which can limit the useful data.

The output of the detector can be expressed by the equation:

$$V_{sig} = K_1 f_1(d)\ f_2\ (t)$$

where K_1 is a constant, f_1 is the modulation of the X-ray signal due to the subject and f_2 is the ripple. The variable _d_ is spatial and _t_ is temporal. In order to make the output a true representation of the modulation function f_1, the equation must have the ripple function eliminated, which is accomplished by dividing the equation by the ripple function $f_2(t)$. It is necessary to generate a signal which has only ripple content and in order to do this, a second detector is incorporated into the system for the purpose of generating the ripple or reference signal. This detector is placed under the subject and the chopper wheel so that the subject function does not appear in its output. Physically, the detector is placed in the side of the lead X-ray cone and penetrates the X-ray beam very slightly so that the chopper wheel is not masked. Figure 3-18 shows the location of the detector with respect to the rest of the components. The output of the reference detector is given by

$$V_{REF} = K_2 f_2(t)$$

K_2 being a constant with constant photomultiplier voltage. In order to divide the signals, the logarithm of each signal is taken electronically and these signals are subtracted. The antilog of the result is then taken, with the result being that the subject signal is divided by the reference signal.

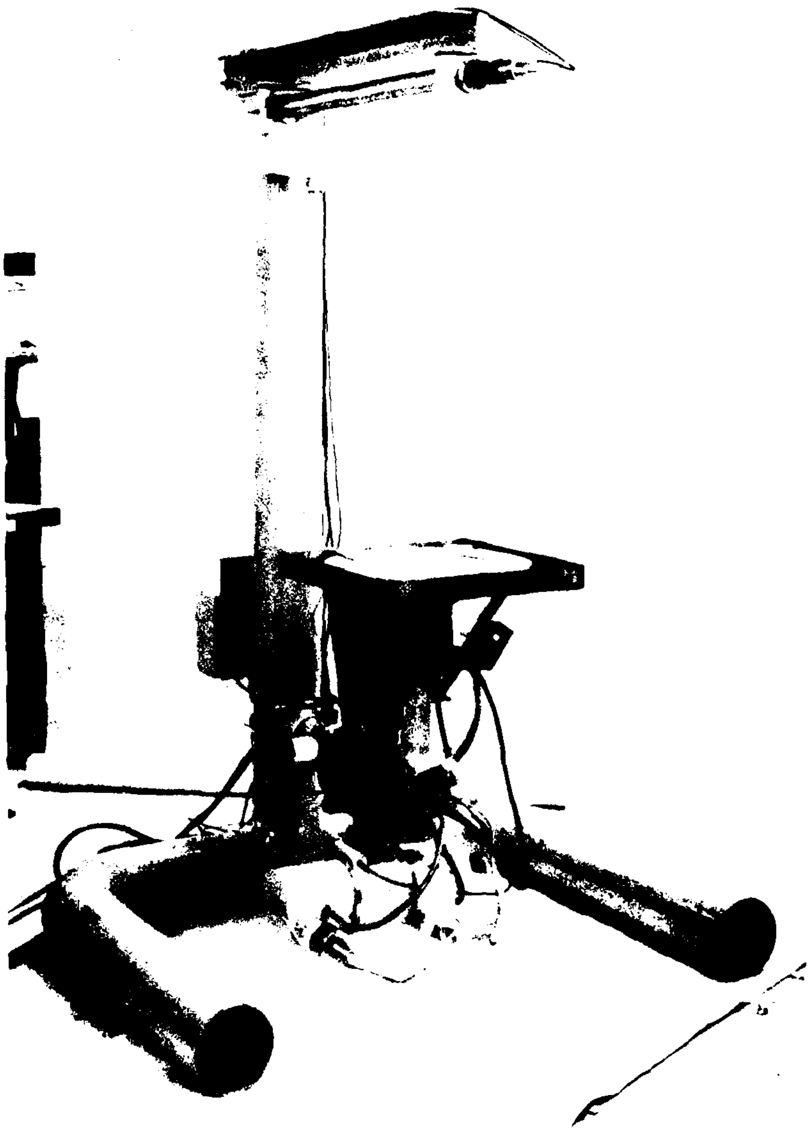

Figure A3-18. Detector

$$\text{Log}^{-1}[(\log (K_1 f_1(d)\ f_2(t))] - \log[(K_2 f_2(t))] = \frac{K_1}{K_2} \times f_1(d)$$

Both constants K1 and K2 are functions of X-ray tube voltage and current and the photomultiplier voltage. The reference detector gain is set at a constant voltage so that finally the only setting which is changed for variations of subject attenuation is the subject detector gain or PMT voltage. This technique is capable of a wide range of tracking in terms of X-ray tube power levels. Figure 3-19 is a diagram of the detection scheme and Figure 3-20 is a schematic of the preamplifier.

The preamplifier operates on the principle that the collector current and emitter-base voltage of a bipolar transistor have a logarithmic relationship for currents between one picoamp and one milliamp. Utilizing this characteristic in the feedback loop of an amplifier, it is possible to construct a linear to logarithmic converter with several decades of dynamic range.

The amplifier IC-1 is a current buffer amplifier, the gain of which is signified by G1. The combination of IC-2 and Q2 as the feedback element provide the actual logarithmic conversion. The second amplifier, IC-3, provides a multiply fraction which in this configuration is a constant, since there is no input to this circuit. Emul is the input voltage to IC-3, determined by the ratio of R8 and R9. The signal at the junction of R12 and R13 is:

$$-\log \text{Emul} + \log E_{REF}\ G_1 = -\log \frac{\text{Emul}}{E_{REF}\ G_1}$$

Gain control is provided by R14 which varies the amplitude of the signal to the base of Q3. The signal on the base of Q3 is given by:

$$-\log E_{SIG}\ G_2 + (N \frac{\text{Emul}}{E_{REF} G_1}) = \ln E_{SIG}\ G_2 \left(\frac{\text{Emul}}{E_{REF} G_1}\right)^N$$

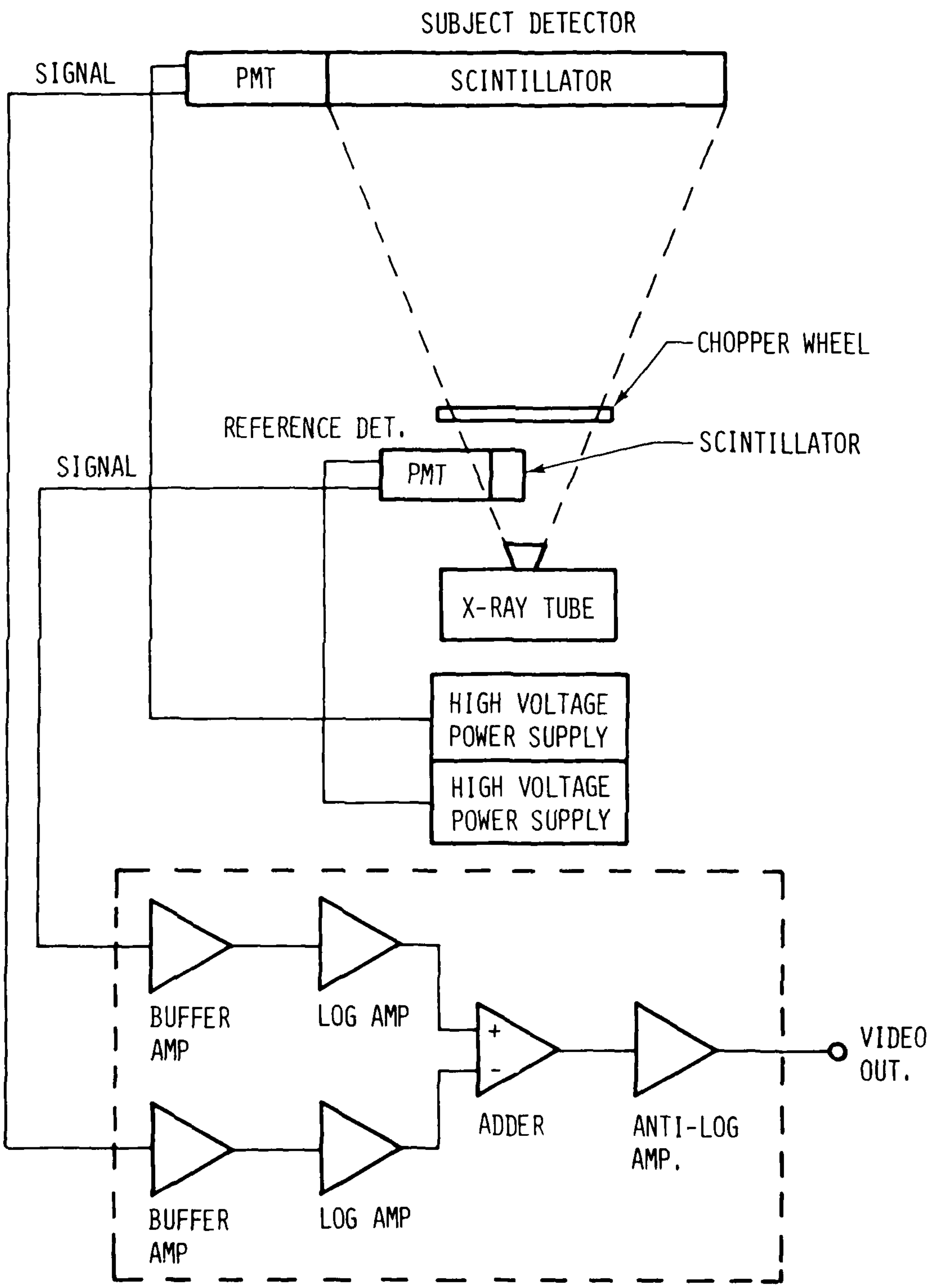

Figure A3-19. Detection Scheme

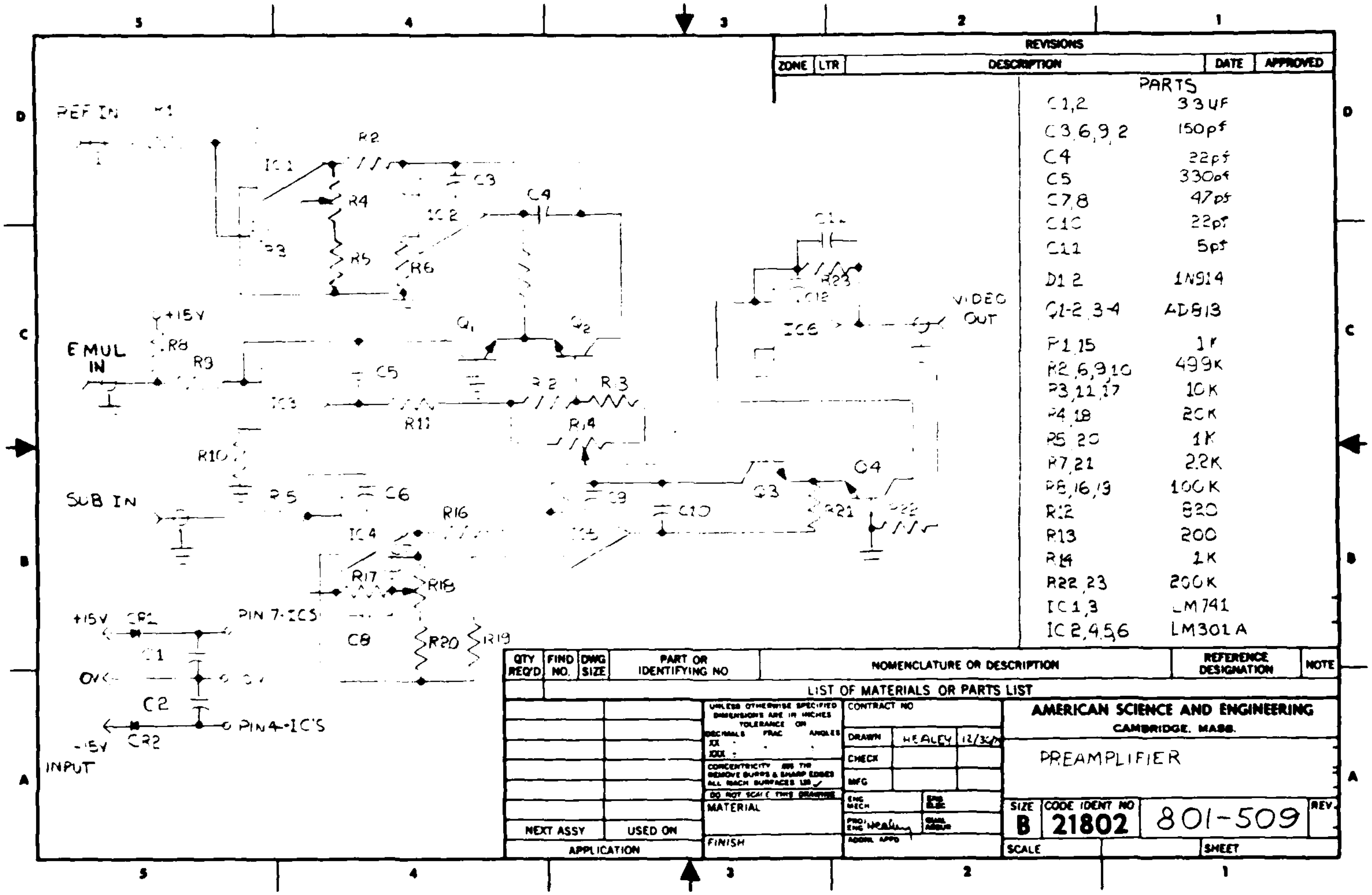

Figure A3-20. Preamplifier

The antilog function is provided by feeding this signal to the emitter of Q4. Since the base of Q4 is grounded, the collector current is the antilog of the signal or voltage impressed on the emitter-base junction. This current is amplified by the amplifier IC-6, the output of which is the corrected video.

Since Emul is a constant, E_{SIG} is effectively divided by $E\ ref^{n}$. The exponent n is controlled by the gain control R14 and is ideally set for n = 1. The variable gain feature allows for some compensation of any differences between the two detectors.

3.7 Display System

The display system consists of the ramp generator, the video storage unit and the television monitor. The display system takes the serial data from the detector together with synchronizing waveforms from the ramp generator and records the X-ray data as a coherent image of the subject. This occurs during the data taking or WRITING mode while the subject is being scanned. Once the X-ray exposure is complete, the system operates in a READ mode where the stored data is displayed continuously on the television monitor until the next exposure. Figure 3-21 is a block diagram of the display system and Figure 3-22 is a photograph of the system together with the photomultiplier power supplies for the detectors.

The subject is scanned in a raster fashion with the rapid scan being across the body and the slow scan being along the length of the body. The fast scan is produced by the action of the chopper wheel and the slow scan is achieved by drawing the subject along the table with a linear motion. The video storage unit requires the video information and two ramp voltage waveforms to store the image properly. The information is written on a storage target in an image format by a scanning electron beam. The two ramp waveforms index the beam and sweep it in a manner which is

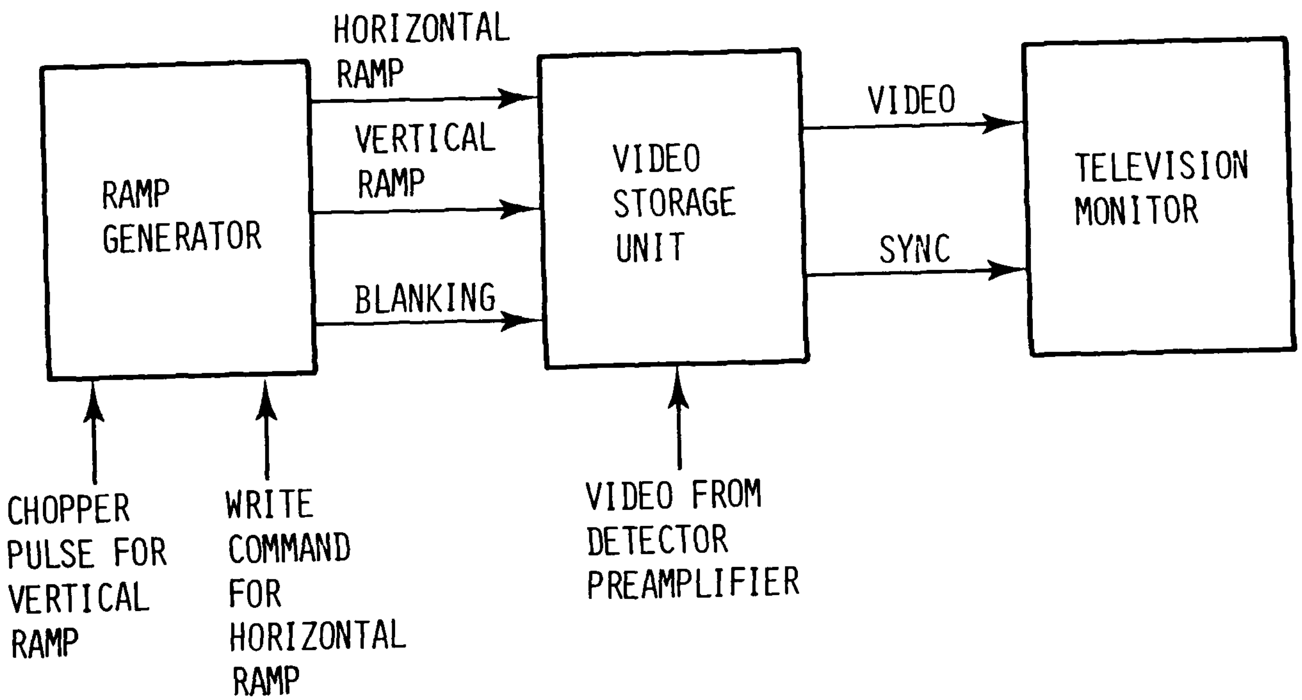

Figure A3-21. Display System Block Diagram

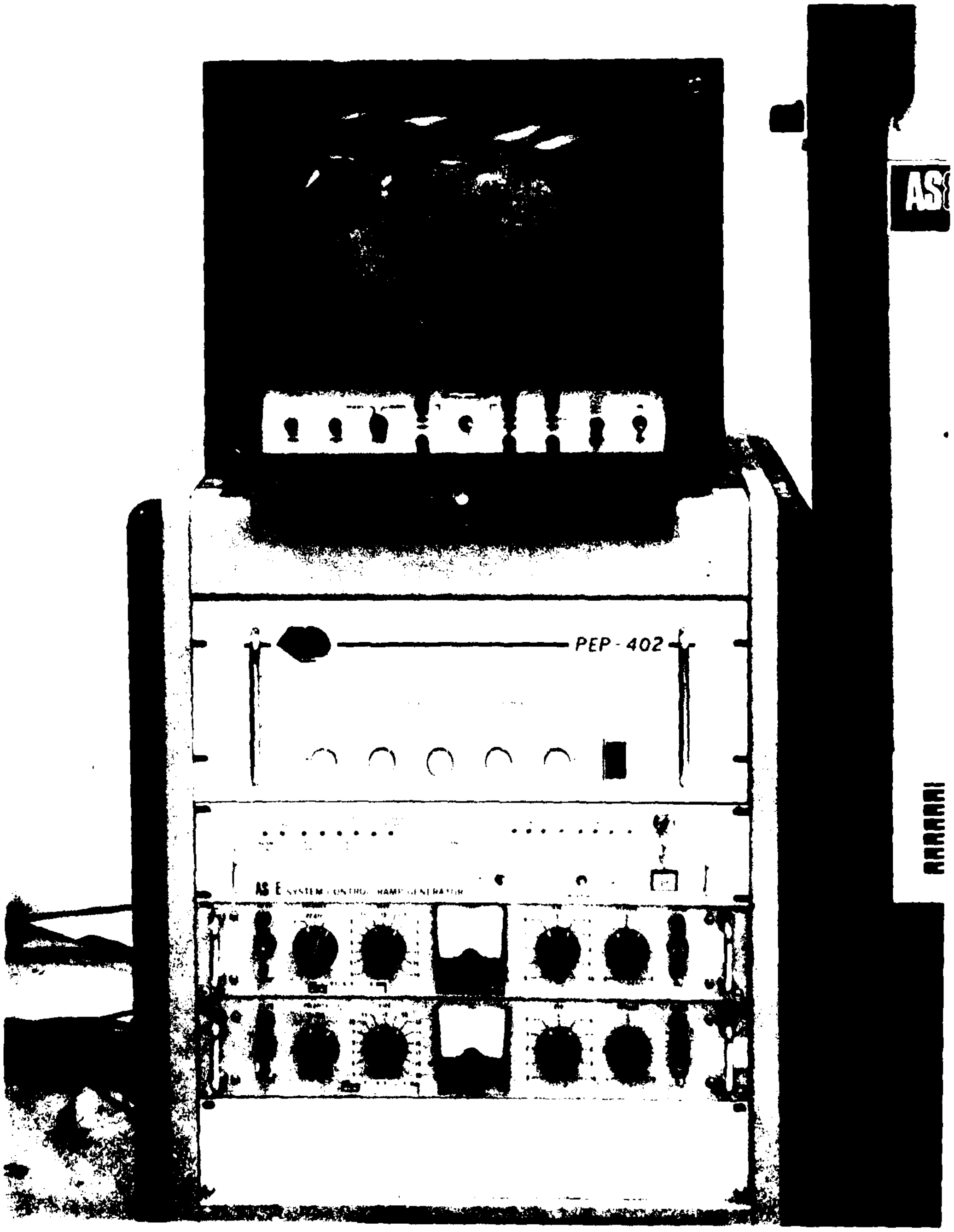

Figure A3-22. System and Photomultiplier Power Supplies

directly analogous to the scanning of the X-ray beam. The amount of current in the electron beam is determined by the amplitude of the video signal so that the resultant image is a stored electron image of the subject X-ray data.

The electron image is stored by virtue of the fact that the target of the storage tube acts like many tiny capacitors in a matrix which is addressed and charged by the electron beam. This information is read by scanning the target with the electron beam in a raster fashion with normal television sweep rates. The beam detects the amount of charge at any particular point on the target and the output signal varies according to the charge density. This signal, combined with normal television synchronizing signals, is fed to the television monitor. The image on the monitor is a direct replica of that stored on the target and therefore, the X-ray image of the subject. The method of reading the target of the storage unit is nondestructive so that the image can be viewed for long periods of time. The reading electron beam detects the presence of the stored charge on the target, but does not change it in the reading mode.

The ramp waveforms control the electron beam in the writing mode and are generated by the ramp generator. The ramp generator requires two signals to initiate the ramps, a slow command and a fast command. The slow command is generated by the control logic when the X-ray tube is energized. This command triggers a slow linear ramp generator which provides a ramp waveform whose duration is exactly the length of the exposure. The voltage varies from -.5 to + .5 volts for a complete sweep of the target in one direction.

The fast waveform is generated from the chopper pickoff signal. The pickoff produces a pulse at the start of each scan of the chopper. This signal is fed to an integrator which converts this

pulse to a ramp voltage. The duration of this ramp is approximately 7.8 milliseconds and the voltage varies from -.5 to + .5 volts in this period.

The video storage unit will only store information when it is in the WRITE mode. In this mode the beam is defected by the ramp waveforms and modulated by the video. The system logic provides a signal which sets the video storage unit in the writing or storage mode. The storage unit is latched in this mode until the X-ray exposure is terminated. At this point the video storage unit returns to a reading mode with all sweep waveforms internally derived. The ramp generator also produces a video blanking signal which, depending on its condition, turns the writing beam on or off in the storage unit. The blanking signal imposes a duty cycle on the video of 1:20 and chops it at a rate of 200 KHz. The video storage unit is only writing on the storage target for 5% of any time period; the rest of the time the beam is cut off by the signal. This is necessary because the length of the X-ray exposure is relatively long and the target is limited as to the amount of charge it can store at any point. The chopping of the video allows the storage tube to be operated in a more favorable and stable region without sacrificing signal to noise ratio. The blanking signal has a protective feature in that the ramp generator will not operate without the chopper pulse. Therefore, if any failure should occur in either the chopper motor or the pick off, the blanking signal will prevent the storage unit from writing in one point of the target and damaging the target.

The target is read in a normal television format in terms of sweep rates. The standard used is a 1023 line, 30 frame per second scan rate. This is an industrial standard and is chosen for equalizing resolutions in the fast and slow scan direction. The normal televison aspect ratio is 3:4, that is, the picture is 3 units high and 4 units wide. If equal vertical and horizontal

resolution is desirable, then there will be 33% more resolution elements in the horizontal direction than in the vertical direction. The scanner operates at 120 scans per second so that in a period of a 10 second exposure, 1200 lines of data will be produced. In order to have equal resolution in the read mode it is necessary to have at least 900 vertical scanning lines. The actual requirement is closer to 1200 scanning lines owing to scanning beam statistics, so that 1023 lines is chosen as the closest commercially standard rate. Also, the 1023 lines are not detectable to the eye when viewing the monitor, which gives a clearer image.

3.8 Mechanical Construction

The main mechanical component of the system is the source/detector keel, shown in Figure 3-18. The chopper assembly is mounted to the lower end of the keel with all cables routed through the center of the support tube. A lead lined enclosure surrounds the source/chopper assembly and this unit in turn slides in place under the transport assembly. The cable route is such that there is no possibility of stray or scattered X-rays getting outside the enclosure. The keel is made of 6" steel tubing and provides a high degree of X-ray shielding. The detector is mounted at the top of the mast and is held by a horizontal support. The X-ray tube cooler is mounted on the mid portion of the mast and is outside the X-ray shielding. The oil hoses for the cooler run through the cable tube with the rest of the cabling.

The transport design takes several factors into consideration. It must be capable of linearly transporting a fairly uniformly distributed weight of at least 200 pounds. It must be transparent to X-rays as the beam must penetrate the transport as well as the subject, and it must have an adjustable drive speed for selective exposures.

The transport frame is constructed of wood and is made in two sections, each approximately 6' long, which slide together. There is a space under the center of the transport to position the X-ray keel and the protective enclosure. The construction is such that the enclosure doors can be removed for access to any part of the source assembly without completely disassembling the transport. A slit 24" x 0.5" allows the X-ray beam to pass out of the source enclosure and through the transport table.

The subject transport is actually a mylar sheet 6' x 3' x 0.010". Mylar was chosen for lowest coefficient of friction when sliding over a flat, wood surface, and because it does not stretch under tensile load. The mylar is fastened at each end to aluminum bars which are, in turn, fastened to a loop of roller chain. The loop of chain runs on rollers on the edge of the table through a protective tube.

The chain is powered by a DC drive motor through a sprocket drive. The drive sprocket and motor are shown in Figure 3-23 with the protective cover removed. The motor is controlled by a continuously variable DC drive located in the main control unit. The table drive is kept centered on the table by a set of nylon rollers on each of the aluminum bars. The material on the table top is oiled birch, which has a very low coefficient of static and sliding friction with mylar.

The drive is car[illegible] e [illegible]ulling a fairly distributed load of 200 pounds with in[illegible]nental [illegible]peed variation of less than 3%. Each end of the table has a double set of interlock switches installed to prevent the transport from being driven beyond the end of the table. The double switches have extensions which contact the transport 2" from the end of the drive so that there is a margin for stopping at the highest drive speed. The switches will also shut off the X-ray power supply in the event that the transport reaches the end of the table.

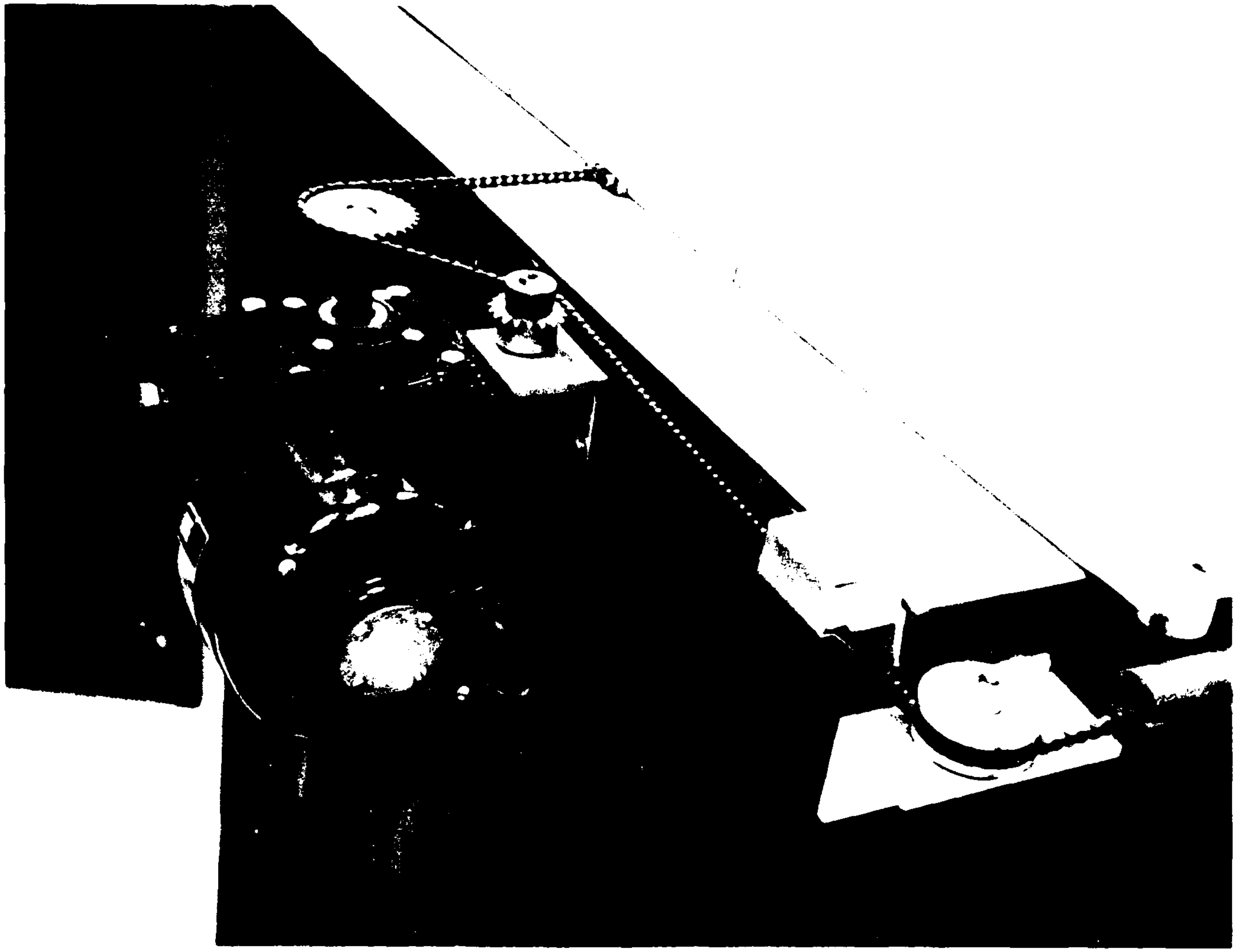

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Figure A3-23. Drive Sprocket and Motor

4.0 OPERATING INSTRUCTIONS

4.1 General

1. Insure that three phase power for the X-ray supply and single phase power for the display system are present

2. Turn on the main breaker and the equipment rack breaker

3. Physically inspect the X-ray cooler on the detector mast to be sure the cooler is running and that the fan has free air circulation

4. Turn on the filament power switch on the filament power supply

5. Set both filament controls to a dial setting of 20 or approximately 1/6 of a turn from the full counterclockwise position

6. Press the filament button on the control panel and observe the current meter. The current should read 3.5 amps. This can be set by the filament standby controls.

7. Wait at least 5 minutes for the tube to warm up.

8. Set the exposure timer to 15 seconds.

9. With the filament control set to the standby position, set the high voltage controller Variac to approximately 1/2 rotation from the full counterclockwise position. This will correspond to a high voltage output of approximately 70KV.

10. Set the voltage and current meter trip needles to the maximum values desired. This is a protective measure for the X-ray tube.

11. Set the table drive to the FWD position and at a speed of at least 16. The transport should be free to travel for at least 10 seconds in that direction. The transport can be positioned by pressing the X-ray/Table Start button and running the drive in the REV direction.

12. Press the X-ray/Table Start button and observe the green HIGH VOLTAGE READY indicator. If it lights, the system is ready.

13. While continuing to press the X-ray button, also press the high voltage ON button and after two seconds the power supply will turn on. At this point the controls will latch until the exposure interval is terminated.

14. Observe the high voltage meter and set the high voltage to the maximum desired value. Generally, a setting of 80 KV. is adequate for initial start-up.

15. Set the tube current with the Filament control to the desired value of 50 milliamperes

16. If time allows, increase the high voltage to 110KV with the control Variac. Continue to observe the current meter as well for any variation in the tube current. If these procedures cannot be completed in one exposure, wait for approximately 1.5 minutes and continue. The button will blink for approximately two minutes following any exposure as a reminder. Do not exceed the heat load of the target under any circumstances. Heat load may be computed by the formula H.U. = Kilovolts x Milliamperes x Exposure Time (Sec) x 1.35. The Dynamax HE67 has a maximum target head storage capacity of 300,000 heat units. At initial start up it is best not to exceed 150,000 heat units

4.2 Display System Check

1. Turn the monitor brightness up until the screen brightens to reveal either a retained image or the target structure. The READ control of the storage unit should be at its index position. If no image is present, rotate this control counterclockwise.

2. Momentarily, turn off the filament supply. When the supply is turned back on, small bright spots will appear on the monitor for approximately 10-15 seconds and then disappear. This indicates that all components are functioning.

3. Set the lower (reference detector) power supply to 660 volts.

4. Set the upper (subject detector) supply to the desired voltage for the type of exposure required. Table 4-1 is a table of the typical detector gain settings for various parts of the body as well as settings for large area scans.

5. Place a piece of lead at least 1/8" thick on the table so that it will pass through the X-ray beam. Take a trial exposure and observe the TV monitor. The lead should appear as dark as the area bordering the light portion of the TV screen. The monitor should be underscanned for this procedure. The blackness can be adjusted by changing the G1 setting of the storage unit. Simply connect an oscilloscope to the G1 pin and observe the voltage level during the write mode. Normally during this mode the black level will be approximately -20VDC. Increasing this value in the negative direction will make the picture appear darker. The lead should be equivalent to the darkest portion of the picture.

4.3 Continuing Operation

1. The system is activated normally by the reflector on the edge of the mylar transport. When the reflector passes under the light beam, the tube will be energized if the table is being driven in the FWD direction with all other conditions met and the speed set-

ting at least 15. The speed corresponding to the dial setting may be found from Figure 4-1. Allow additional space for the start up of the X-ray tube, which takes two seconds following the trigger from the photodetector. Placement of the reflector allows for accurate and repeatable exposures over the desired area of the body.

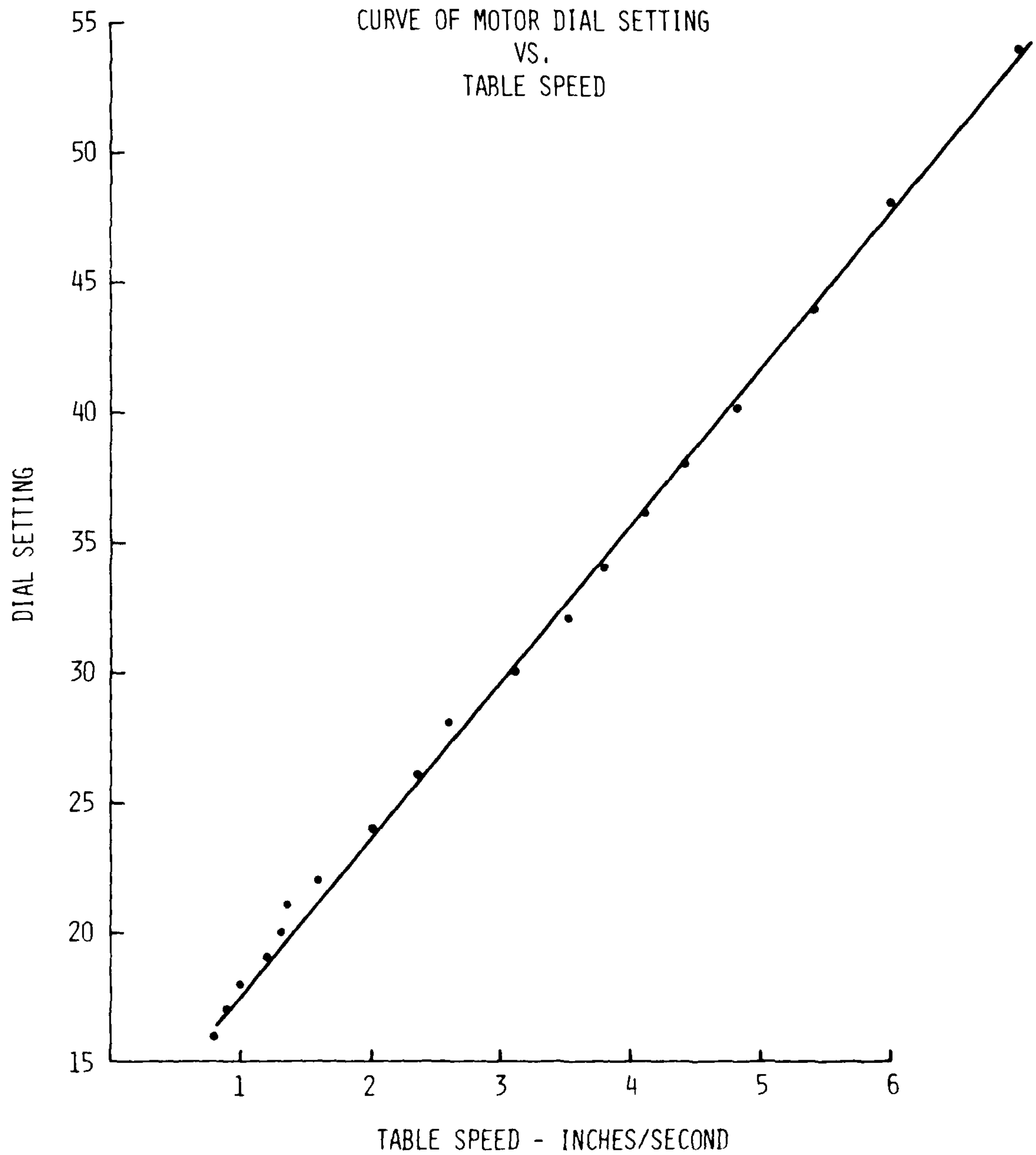
CURVE OF MOTOR DIAL SETTING
VS.
TABLE SPEED
DIAL SETTING
TABLE SPEED - INCHES/SECOND
55
50
45
40
35
30
25
20
15
1
2
3
4
5
6

Figure A4-1

2. The gain of the subject detector is controlled by the upper high voltage power supply. Table 4-1 gives typical setting for optimal exposures. These values were obtained from an X-ray phantom and may vary slightly from subject to subject.

Table 4-1

Typical X-Ray Exposure Parameters

Area	Drive Speed in./sec.	Detector Voltage	X-ray Tube Volts (KV)/Current (MA)
Head	1.0	850	110/50
Chest	1.33	700	110/50
Pelvis	1.33	850	110/50
Thigh	1.33	700	110/50
Leg (thigh included)	2.35	770	110/50
Foot	1.0	700	110/50
Arm	1.6	650	110/50
Hand	0.8	550	110/50
Hand	0.8	590	70/50

5.0 SYSTEM PERFORMANCE

5.1 Stray Radiation

The body scanner was surveyed for stray radiation in the area surrounding the X-ray generator. Except for the scanning beam port, the stray radiation measured was only twice that of the natural background radiation.

5.2 Radiation Doses

Radiation exposures to various parts of the body were measured using a RANDO X-ray Phantom to simulate the body. The phantom is approximately tissue equivalent and has about the same response to X-rays as the human body. Measurements were made by a certified health physicist using thermoluminescent dosemeters (CaF) and read on a Harshaw 2000 series system. Calibration of the measurement technique was by means of a cross calibration against National Bureau of Standards certified tissue equivalent ionization chambers utilizing constant potential X-rays at energies comparable to those used in the scanner. All data was taken at 110 KV and 50 ma, with a transport speed of 1.2 inches per second. Measurements were made on the various parts of the phantom so that back scatter and dose build-up are typically represented.

MEASUREMENT LOCATION	DOSE PER EXPOSURE
Chest entry	323 microroentgen
Chest exit	25 microroentgen
Pelvis Entry	247 microroentgen
Mid-torso (inside phantom)	72 microroentgen
Pelvis Exit	25 microroentgen
Alternate chest entry	265 microroentgen

All measurements are accurate to ± 3 microroentgens.

5.3 Resolution Test

The resolution capability of the body scanner was determined by using a Nuclear Associates, Inc. X-ray test pattern. The pattern is a variable spacing bar chart made of a 50 micron thick lead encased in plastic. The test pattern has resolution groups which are separated by index marks. The resolution of each group, starting from the left side of Figure 5-1 is:

Group	Resolution lp/mm
1	0.25
2	0.5
3	0.6
4	0.7
5	0.85
6	1.0
7	1.2
8	1.4

Figures 5-1 and 5-2 are radiographs of the test pattern taken with a Tektronix type 602 Display Unit with a Type 50/70 Scope Camera. This technique provides the best means of reproducing photographically the quality of the image on the television display. The type of film used is Polaroid type 105 P/N which gives both a positive and a negative within one minute of exposure. Figure 5-2 shows the pattern with the bars normal to the fast scan and Figure 5-1 shows tha bars parallel to the fast scan. Clearly, the 0.7lp/mm group is visible, which shows that the X-ray beam is able to differentiate between two lines with a spacing of 0.7mm. Figures 5-3 and 5-4 show the same test pattern as seen through 8 inches of water. The photomultiplier gain was increased to obtain these photos.

5.4 Representative Images

Table 5-1 is a list of representative examinations conducted with the laboratory X-ray scanner. Images were recorded using Polaroid film viewing a slow scan cathode-ray tube display as described

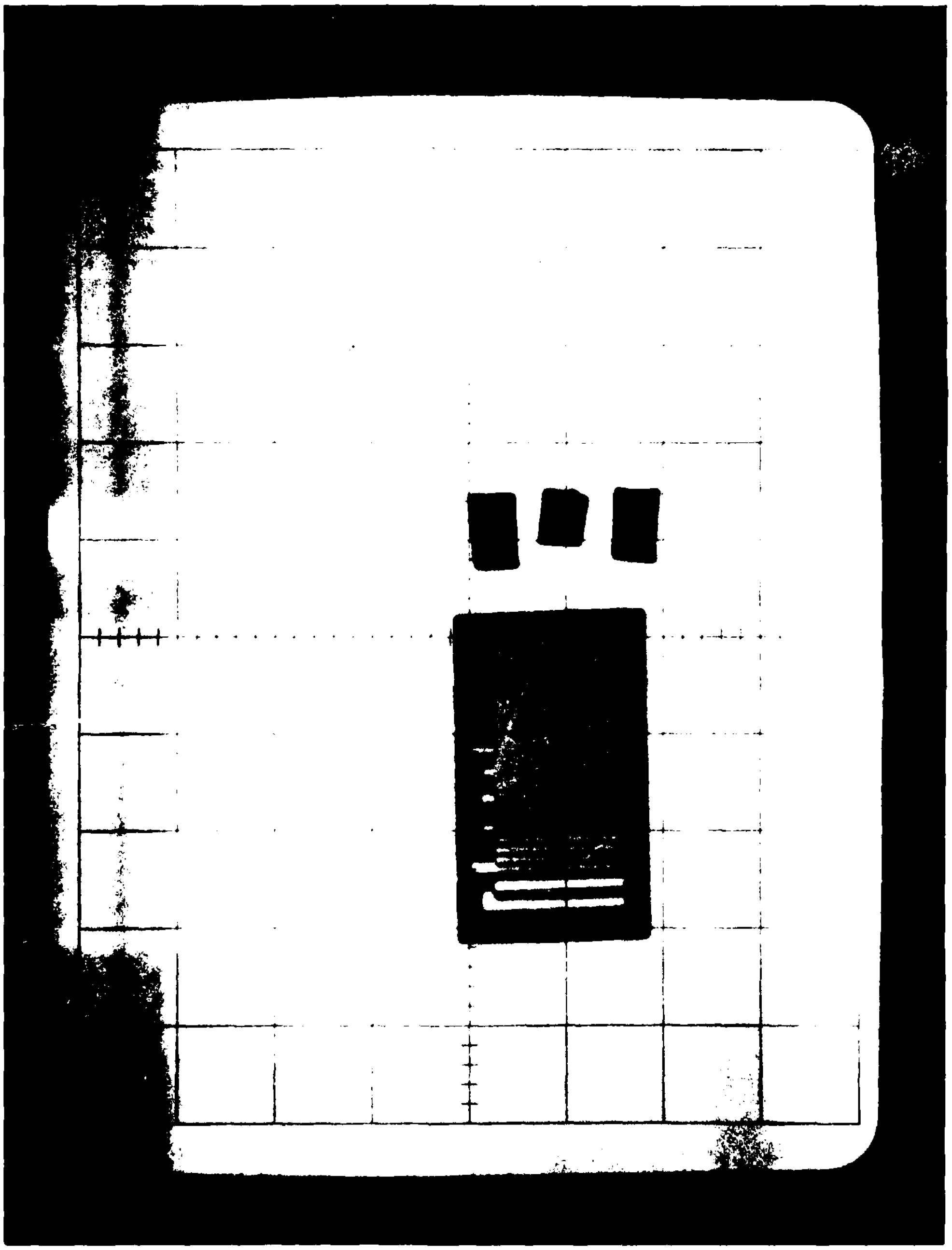

Figure A5-1. Radiograph of Test Pattern

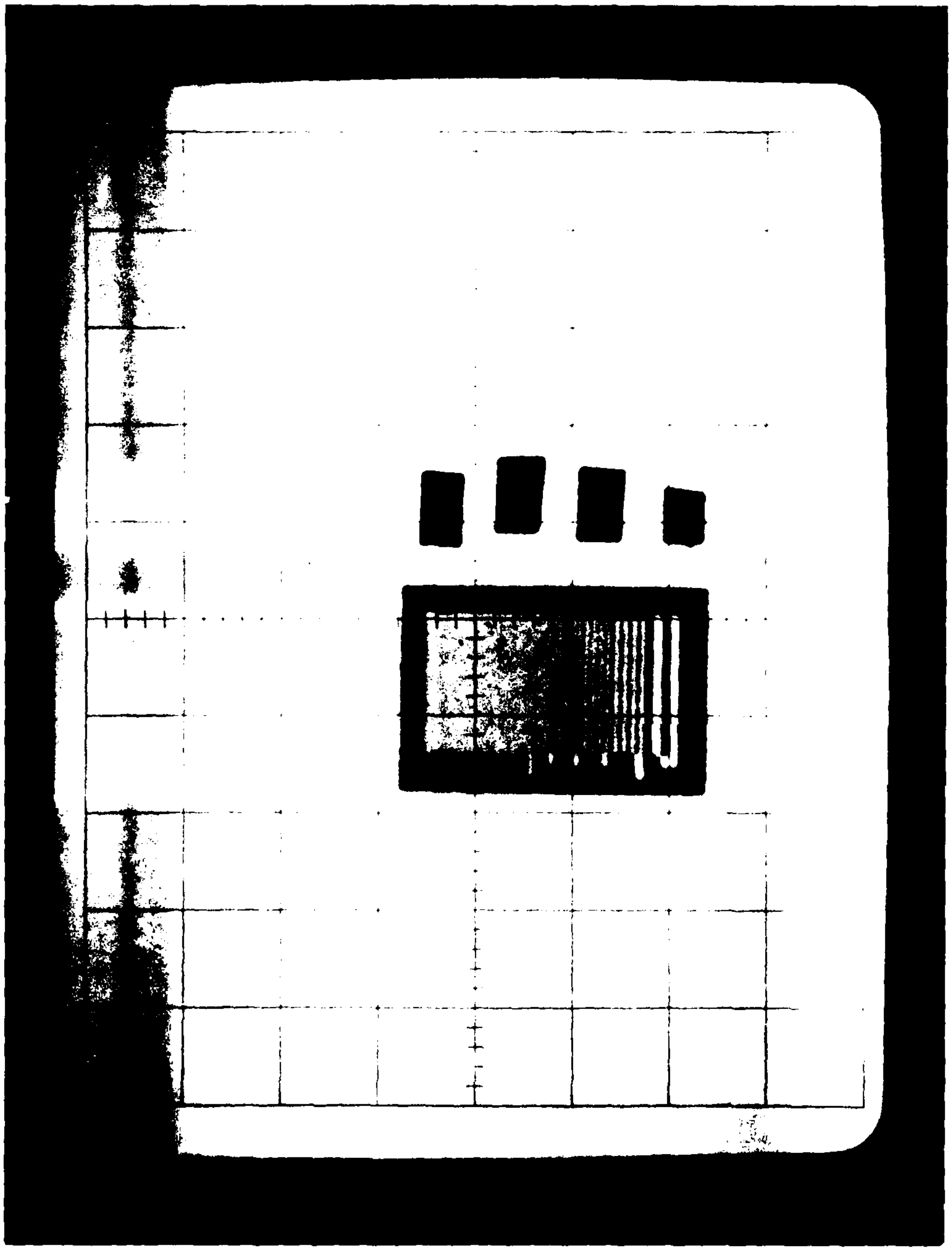

Figure A5-2. Radiograph of Test Pattern

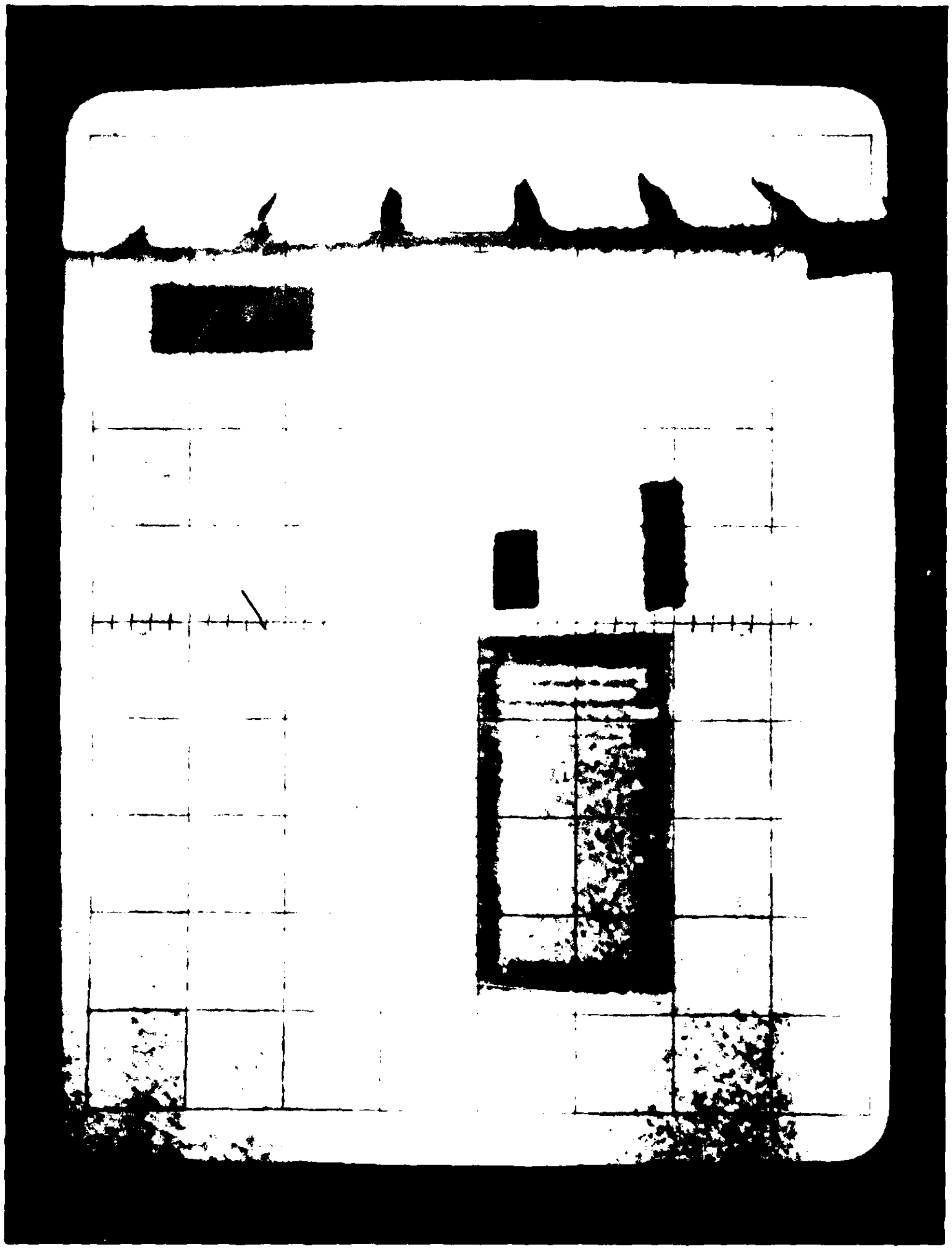

Figure A5-3. Radiograph of Test Pattern

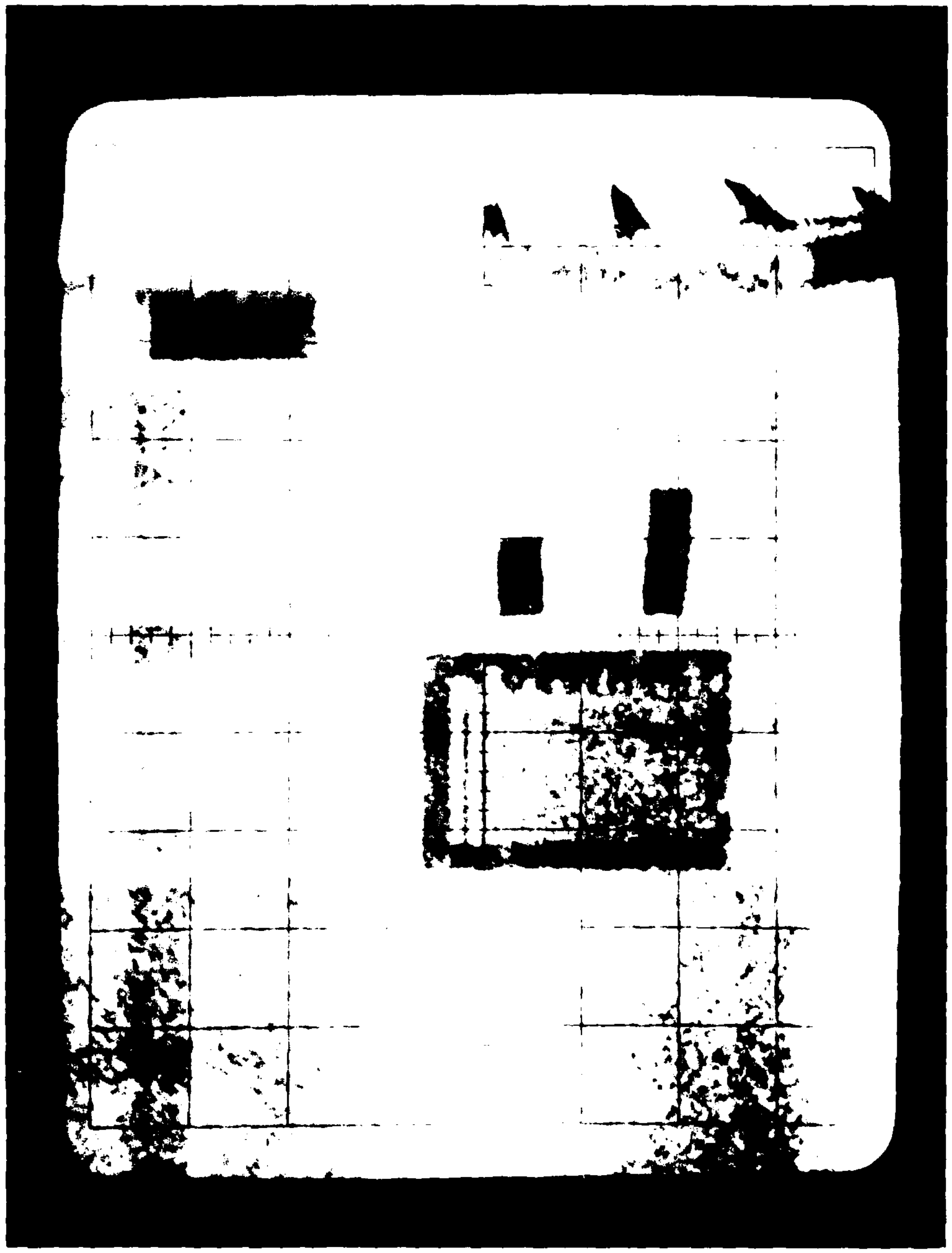

Figure A5-4. Radiograph of Test Pattern

above Figures 5-5 to 5-14 show the X-ray images corresponding to each entry in Table 5-1.

A distinctive feature of these images is the clarity with which small metallic objects are discriminated. The #6 steel nuts and screws with typical smallest dimensions of a few millimeters are all readily discerned. Further, a geat deal of additional radiological information is available in these images, the value of which requires a much more extensive clinical evaluation.

TABLE A5-1

Figure	Subject	X-Ray Tube Voltage	PMT Voltage	Transport Speed	Exposure Time	Comments
5-5	Skull, cervical and thoracic vertebrae	110KV @ 50MA	850	1.0 in/sec	15 sec.	Circular objects are #6 steel nuts, 5mm O.D., 2mm I.D.
5-6	Same as 5-5	110KV @ 50MA	850	1.0 in/sec	15 sec.	Photo taken from storage display TV monitor.
5-7	Shoulders and Chest	110KV @ 50 MA	700	1.33 in/sec	15 sec.	Objects are #6 steel screw, 10 mm long and #6 steel nut.
5-8	Shoulders and Chest	110KV @ 50 MA	700	1.33 in/sec	15 sec.	Photo taken from storage display TV monitor.
5-9	Pelvis	110 KV @ 50MA	850	1.2 in/sec	15 sec.	Objects in upper portion of picture are #6 steel screw and nut.
5-10	Pelvis and Thighs	110KV @ 50MA	850	1.4 in/sec	15 sec.	"
5-11	Knees and Lower Legs	110KV @ 50MA	770	2.35 in/sec	15 sec.	"
5-12	Foot	110KV @ 50MA	700	0.8 in/sec	15 sec.	"
5-13	Hand and forearm	110KV @ 50 MA	650	1.3 in/sec	15 sec.	Object is #6 steel nut.
5-14	Hand	110KV @ 50MA	650	0.8 in/sec	15 sec.	"

Figure A5-5. X-ray Image of Skull, Cervical, and Thoracic Vertebrae

AD-A100 366 AMERICAN SCIENCE AND ENGINEERING INC CAMBRIDGE MA F/G 6/5
DEVELOPMENT OF A WHOLE BODY FLYING-SPOT X-RAY MEDICAL UNIT.(U)
JAN 80 J A STEIN DAMD17-74-C-4071
UNCLASSIFIED ASE-4646 NL

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Figure A5-6. X-ray Image of Skull, Cervical, and Thoracic Vertebrae from Storage Display T.V. Monitor

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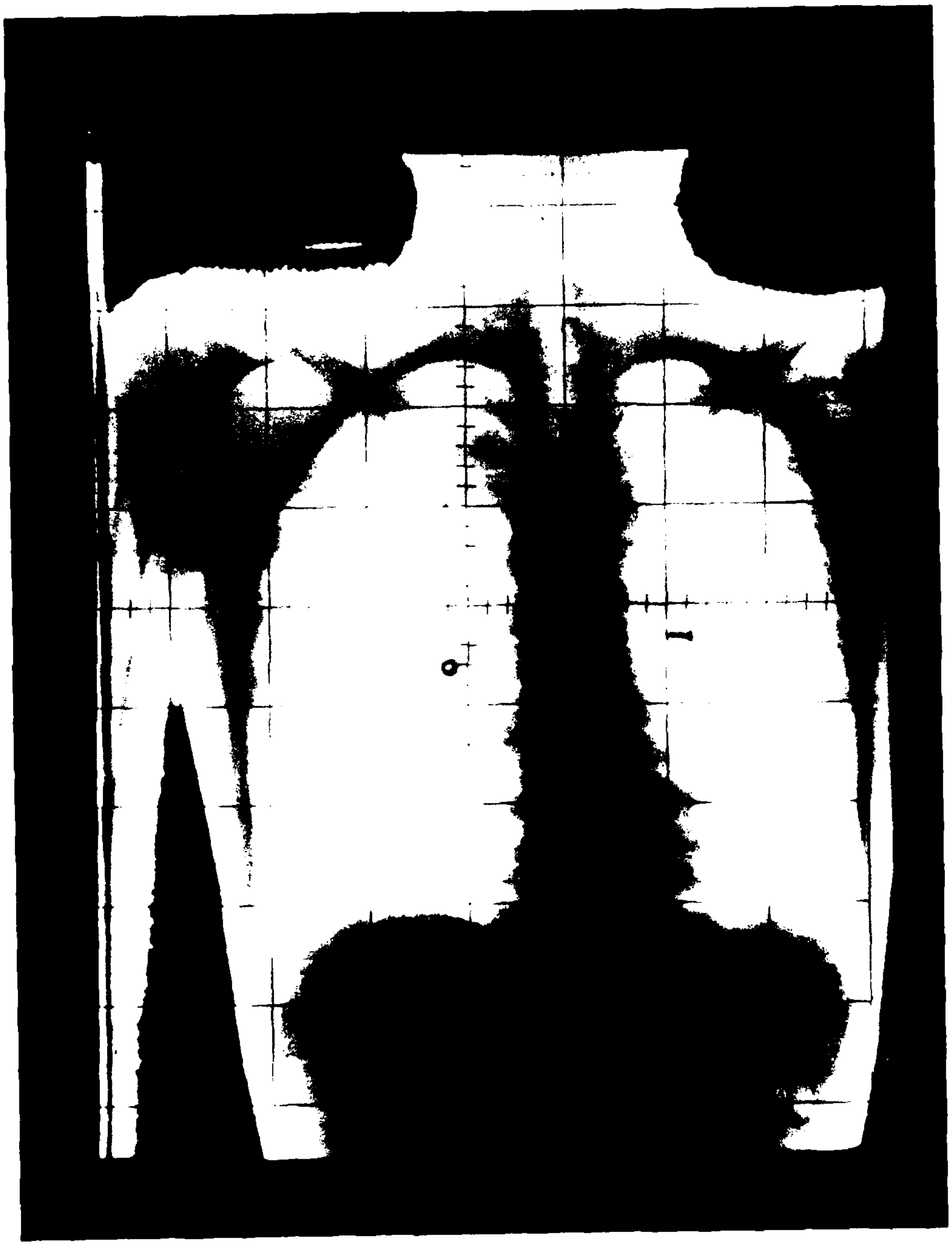

Figure A5-7. X-ray Image of Shoulders and Chest

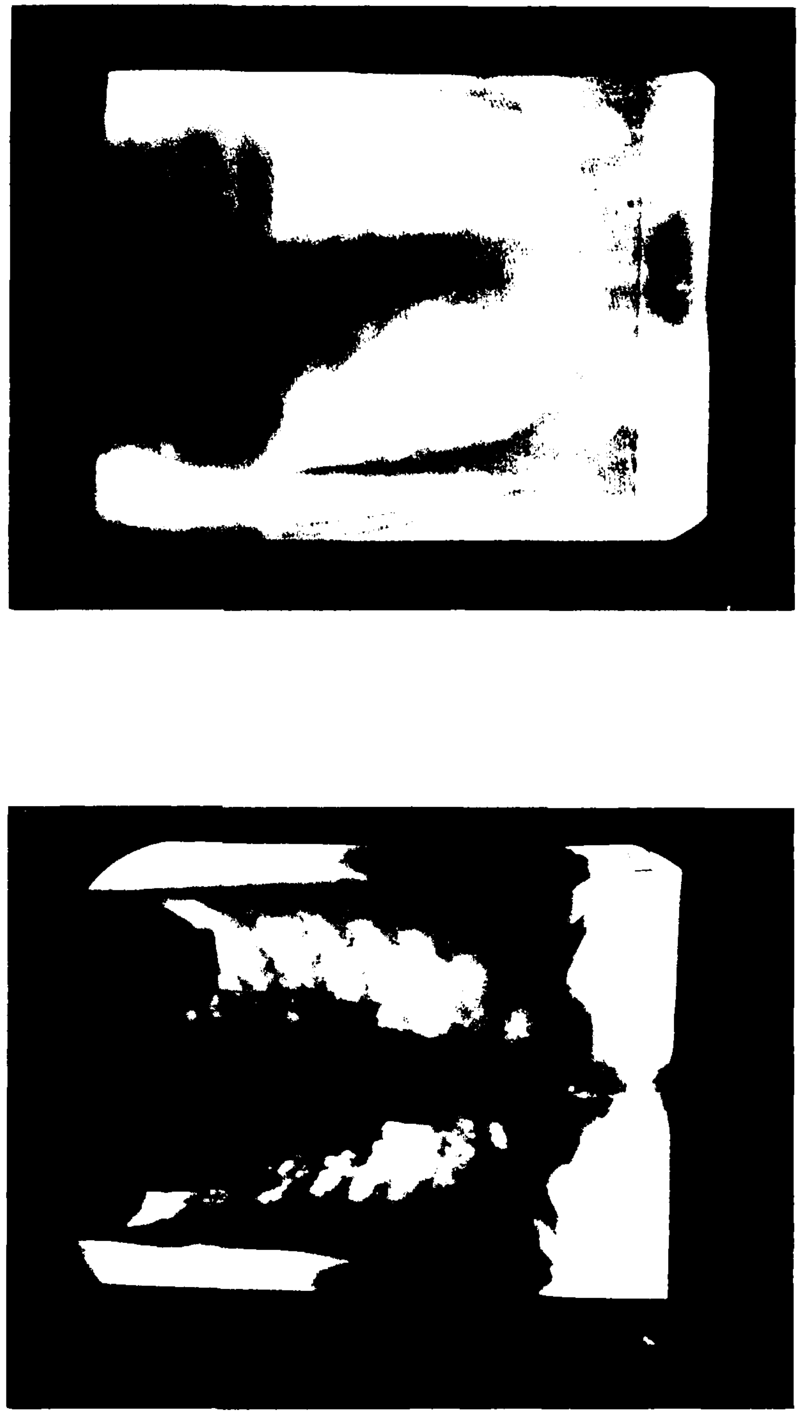

Figure A5-8. X-ray Image of Shoulders and Chest from Storage Display T.V. Monitor

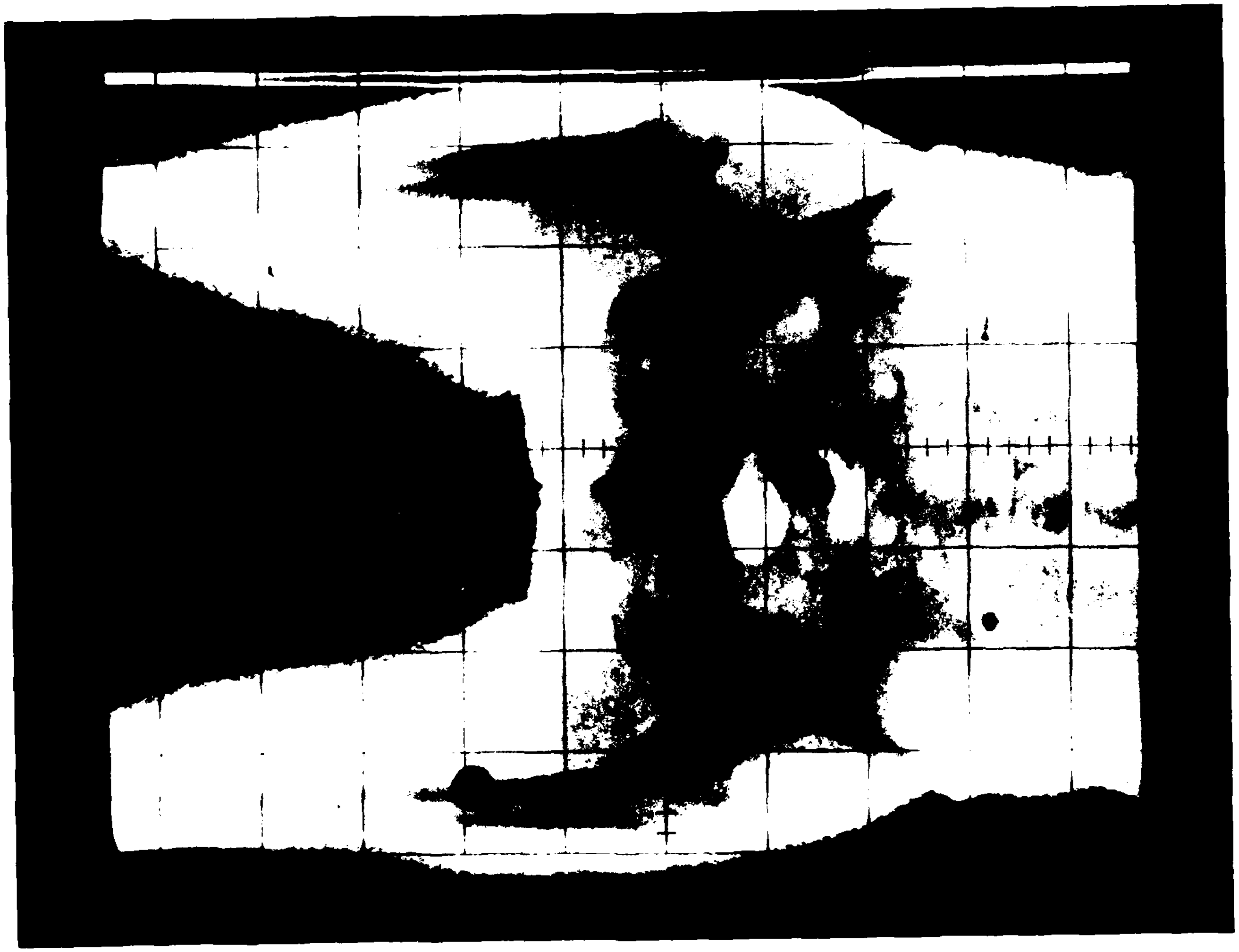

Figure A5-9. X-ray Image of Pelvis

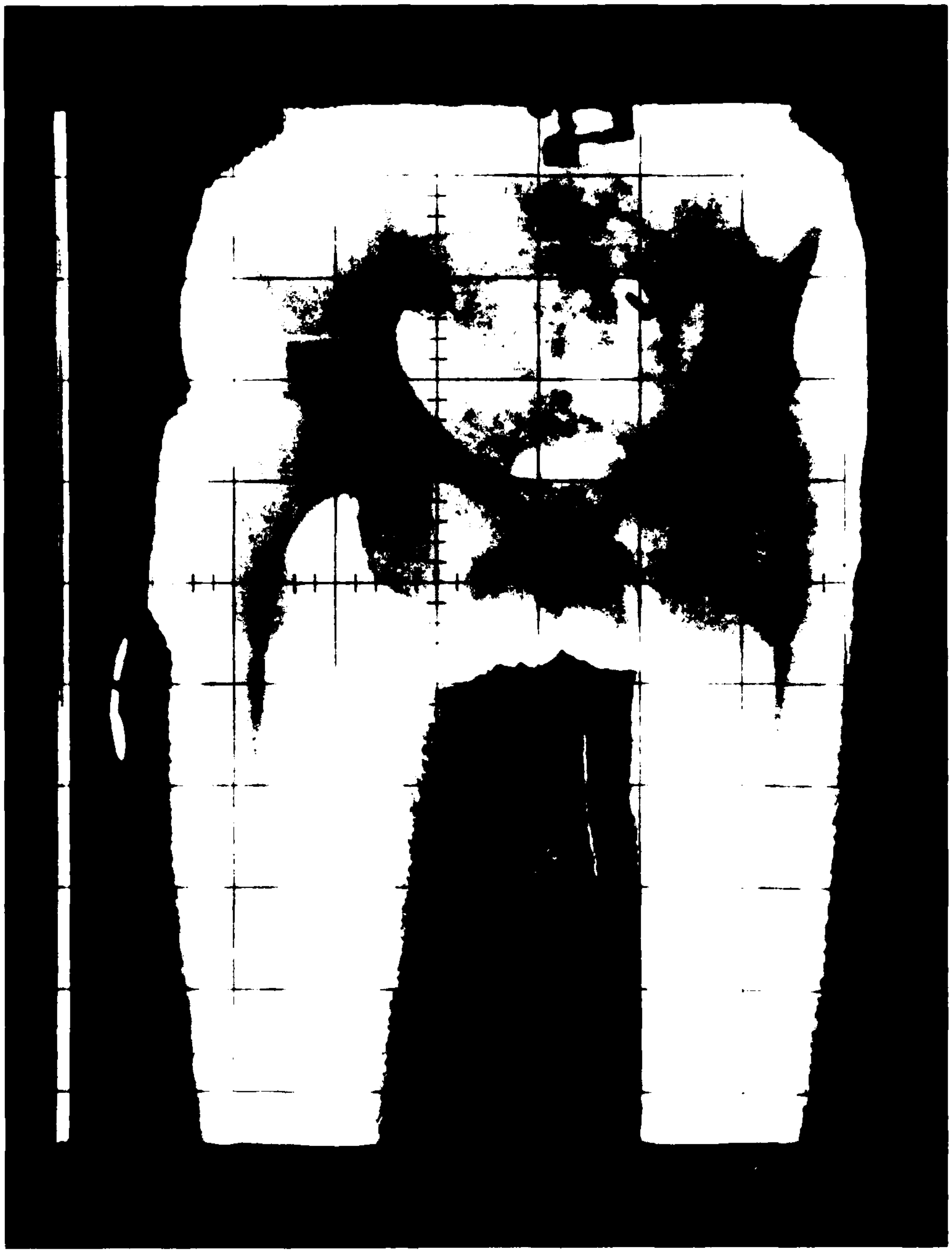

Figure A5-10. X-ray Image of Pelvis and Thighs


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Figure A5-11. X-ray Image of Knees and Lower Legs

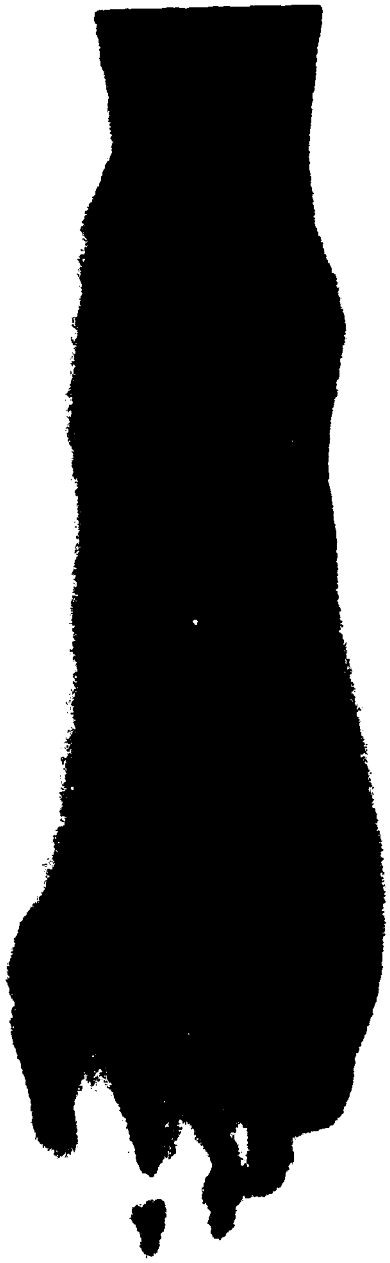

Figure A5-12. X-ray Image of Foot

FP-754

Figure A5-13. X-ray Image of Hand and Forearm

Figure A5-14.   X-ray Image of Hand

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